Modeling and Analysis of Thermal Loading and Heat Transport in Restored Teeth

Doret Spierings

MODELING AND ANALYSIS OF THERMAL LOADING AND HEAT TRANSPORT IN RESTORED TEETH

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CONTENTS

MODELING AND ANALYSIS OF THERMAL LOADING AND HEAT TRANSPORT IN RESTORED TEETH.

Note: The fact that several chapters will be or be published in separate papers in different journals, made it inevitable that some sections in a number of chapters are repetitive in nature.

GENERAL INTRODUCTION

Thermal shocks in the oral cavity may affect teeth and thereby damage their structures. Temperature changes during dental treatment and subsequent histological changes of the pulp tissue have been widely reported in the literature (Zach 1962; Kirschner 1984; Walther 1984). In general, a restorative procedure leading to the substitution of decayed tooth structures by restorative materials (amalgam), results in an enlarged thermal diffusivity of the restored teeth. Especially, the restoration-dentin interface will show an increased temperature gradient (Voth 1966; Tibbetts 1976). However, the resulting residual changes in temperature distribution within the tooth structures are not known. Thus far, the attention has been focused on temperature measurements and effects resulting from various thermal loading conditions (Trowbridge 1980; Augsburger 1981). This thesis deals with the underlying thermal phenomena, such as thermal diffusivity and temperature distribution within teeth induced by a certain thermal load. Especially, the influence of restorative materials and the size of a restoration on the temperature distribution within a tooth will be of major concern.

Healthy unrestored teeth may be affected by alternately normal and extreme thermal loading. For example, cracks and propagation of cracks in the enamel have been reported (Brown 1972; Lloyd 1978a). Fractures in the tooth structure may also be induced by dry cutting during restorative procedures (Brown 1978). Burn lesions in dentin and affected pulp tissue as results of dental treatment have been observed despite the presence of a cooling device (Langeland 1960, 1970).

To cure composite resin restorative materials, an exothermal reaction takes place (Plant 1974, Bausch 1982). It is assumed that such thermal stimuli are responsible for postoperative thermal discomfort in 38 to 50% of the patients having had restorative treatment (Silvestri 1977; Piperno 1982; Miller 1984). However, a recovery of thermal complaints of patients within 1 week to 6 months has also been reported (Piperno 1982). Although the complaints of thermal stimuli diminish in the long-term, fractures and propagation of microcracks might occur in restorative materials (Roydhouse 1970; Williams 1983; Montes-G 1983) causing microleakage (Peterson 1966; Munksgaard 1985). These and other thermal studies are more extensively reviewed in chapter two.

In order to study heat transport within teeth, in vivo or in vitro experiments are the methods of choice. These direct experimental methods are however limiting heat transport research to temperature measurements within teeth. In vivo experiments are for ethical reasons not suitable to analyze thermal changes in the pulp or in its vicinity. Moreover, the enormous biological variation in experimental conditions do not allow for easy generalizations based on a small number of individual observations.

Mathematical methods allow the study of the transient temperature field within a theoretical model of a tooth. Calculations are carried out using Finite Element Analysis (FEA). This numerical method has been used earlier to analyze thermal stresses in an axisymmetric tooth model (Brown 1978; Lloyd 1978a). Takahashi (1982), using the same method, analyzed the intrapulpal temperature change of a two-dimensional tooth model under a steady-state loading condition. The aforementioned studies focused on steady-state problems and thermal effects. In contrast to a steady-state problem, the research reported in this thesis is directed towards <u>transient</u> thermal processes.

Using a numerical mathematical method creates problems at a different level as compared to experimental research. If, for instance, knowledge concerning specific input data is insufficient, assumptions have to be introduced inevitably. Such assumptions will finally limit the generalization of results. If the modeling chosen appears to be strongly influenced by one or more of the assumptions, supplementary experiments may be necessary.

Numerical mathematical modeling like in this FE-method is very suitable for parametric studies. Parametric studies using FEA may also be indicative for further experimental research. Another advantage of using a numerical mathematical method is the extensive information obtained after calculation of the temperature distribution in teeth, e.g. the temperature within a tooth model as a function of place or time. The FEA opens also the possibility to calculate the influence of various parameters upon the temperature field and extreme clinical thermal conditions. Although, a numerical mathematical method has its own limitations, such a method has to be preferred to gain more insight into the problem to be solved.

The aim of this study is to analyze the heat transmission and temperature distribution within various restored teeth under certain transient thermal loading conditions.

Thermal studies concerning temperature measurements and thermal injuries are more extensively reviewed in chapter two. Section 2.1 deals with the surface temperature of oral tissues, whereas section 2.2 emphasizes various thermal loading conditions in the oral cavity and their after-effect on teeth. In chapters three and four of this thesis the theoretical model will be described and analyzed. Since in the chosen model, assumptions are made concerning two aspects of the thermal loading conditions, supplementary experimental and theoretical research is necessary. The attention will be drawn to: (1) the thermal load in the oral cavity caused by drinking hot/cool liquids; and (2) the value of the convective heat transfer coefficient of this medium. The related experiments will be described in chapter five.

An outline of the methodology is summarized for a better understanding of the inter-relations between the sections of chapter five:

Part A (section 5.4) comprises a mathematical model based on FEA in which the following factors can be varied: the geometry of the tooth (axisymmetric); the shape of the restoration; the material properties (thermal diffusivity); the convective heat transfer coefficient (htc); and the thermal loading. The modeling chosen results in output data like "the temperature as a function of time" at any point within the tooth model or results in "the temperature as a function of place" at any time. This is illustrated in chapters three and four.

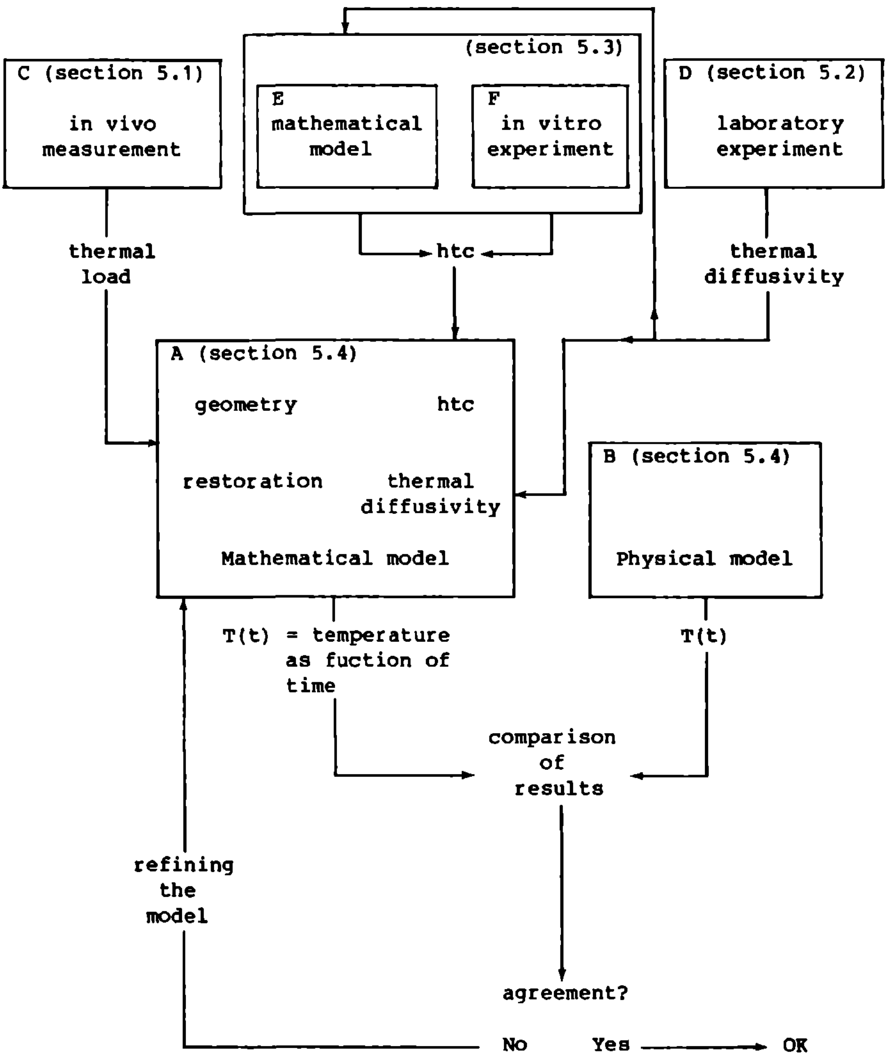
Part B (section 5.4) describes a physical model, i.e. an axisymmetrical resin tooth model, situated in the oral cavity of a person. The geometry equals a specified realization of the mathematical model. The thermal diffusivity of the resin used is determined by separate measurements (see part D). The thermal loading is known (see part C) resulting in a temperature change within the model as recorded by a thermoelement.

Part C (section 5.1) deals with an in vivo experiment in which the temperature near the tooth was recorded over time durig the drinking of hot/cool liquids. The given temperature-time curve is described with a mathematical equation. Thus, the derived function may serve as input for the model of part A.

Part D (section 5.2): the thermal diffusivity of the resin used in the physical models of part B and F is determined in a laboratory experiment using a stepwise temperature change. The experimentally determined value of the thermal diffusivity will be used in the simulation models of part E and A.

Parts E and F (section 5.3) describe a mathematical model and an in vitro experiment. Together, they are used to determine the value of the heat transfer coefficient (htc). Part E is the mathematical model of the tooth in part F, subjected to a simple thermal load. Part F is therefore the physical experiment described by part E. In part E, the value htc is chosen such that the calculated results agree with the recordings of F. The value determined for htc will be used in part A. As far as necessary the value of the thermal diffusivity determined in part D is used in parts E and F.

The main goal is to develop a theoretical model (A) which is a good and realistic schematization of the physical model (B). If so, the mathematical model is valid for future studies. If not, the mathematical model needs further refinements.



An outline of the methodology of chapter five

2.1 SURFACE TEMPERATURE OF ORAL TISSUES*

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ABSTRACT - The storage of heat in the human body is regulated by a meticulous physiological control of heat production and heat loss. The heat regulation of the dentition and in particular of the pulp tissue is still disputed.

Several methods have been used for measuring the surface temperature of the oral tissues. Until now a thermocouple seems to be the easiest instrument to use for this purpose. Although there is a lack of information about environmental and testing conditions, an approximation of the surface temperatures of the dentition (30-35^{o}C) and soft tissues (32-37^{o}C) has been made based on a literature review.

INTRODUCTION

It is well known that continuous irritation caused by thermal fluctuations or exposure to an extreme thermal shock will result in damage of pulp tissue (Lisanti 1952; Postle 1959; Zach 1965; Robinson 1962). The percentage of dental treatment resulting from thermal overloading has not been investigated epidemiologically until now. However, the greater part of extractions and endodontic treatments can be considered to be the result of carious lesions and/or subsequent restorative procedures. During restorative treatment, irreversible alterations are frequently caused by thermal loads (Robinson 1962; Nyborg 1968; Langeland 1970).

To gain insight into thermal processes within the oral cavity, the attention is drawn to the system of heat regulation. A review will be given of recent literature with respect to surface temperatures at different sites in the oral cavity. The methods used in the various studies are discussed and on the basis of this, a mean surface temperature for soft tissues and teeth is chosen.

HEAT REGULATION

The law of conservation of energy is applicable to each system of heat regulation. In the case of the human body it means that the amount of heat intake and heat production (as a result of food metabolism) is equal to the total amount of heat loss and accumulated heat. The storage of heat in the human body is the result of two main processes: heat production and heat loss. Both processes are regulated by meticulous physiological means.

Heat production

Heat production is a continuous process of transforming the latent chemical energy of food into heat. The daily intake of food guarantees a constant supply of energy, enabling the human body to a constant heat emission. Moreover, heat production supports metabolic processes, which in turn are responsible for the heat production as well. These processes therefore form a vicious circle as part of a system that keeps the body at a nearly uniform thermal level in spite of external influences.

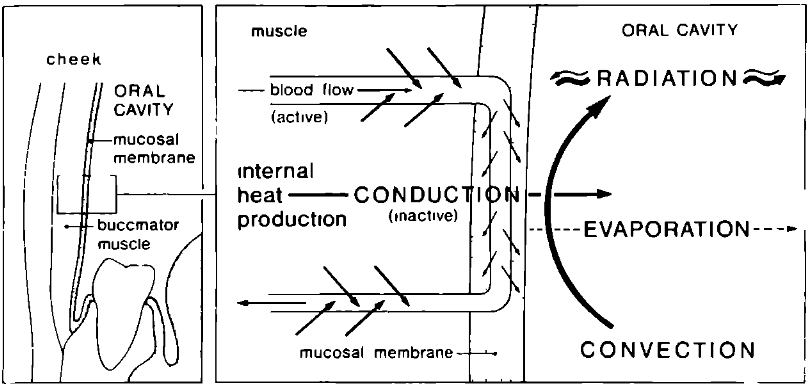

Fig 1 - Schematic representation of heat transmission in the oral cavity.

Heat loss

Heat loss mainly takes place through the skin surface and to a limited extent via the mucous membranes of the oral cavity and the respiratory tract. Two mechanisms are involved in the release of heat by the body: a physical and a chemical one.

The physical heat loss consists of a complex mechanism, which can be simplified by the following four processes: a) radiation; b) convection; c) conduction; and d) evaporation. These processes are schematically illustrated in Figure 1. Activation of the muscles of the jaw for example will result in internal heat production, starting a passive heat transfer by heat conduction. At the same time the blood circulation in the tissues contributes to an active heat transfer. At the skin surface and at the mucous membrane the heat loss mainly takes place by radiation and convection. Only 25 % of the total heat loss is the result of evaporation. Evaporation plays a much greater role when there is convection at the surface.

At extreme low environmental temperatures heat loss of the body is minimal. However, the physical mechanism of heat production is then no longer able to maintain a uniform body temperature. The additional activation of a chemical mechanism defends the body temperature from a decrease. Exothermal chemical reactions in metabolic processes become responsible for the greater part of the heat supply.

A precise regulation of heat production and heat loss is essential for optimal functioning of the human body as a biological system. The system maintains a constant body temperature of approximately 36.5^{o}C, which in essence is the temperature of the so called central core. Regulation of the core temperature is transmitted via thermoreceptors in the hypothalamus.

Heat regulation of the dentition and in particular of the pulp tissue is still disputed. Are there peripheral thermoreceptors in human teeth? Naylor (1962; 1964) has shown that experimental subjects react to thermal stimuli applied to the dentition. The causal source however could not be distinguished from other stimuli. Trowbridge (1980) attributed the mechanism of reaction to thermal stimulation to hydrodynamic forces in the dentinal tubules. Recently, Jyväsjärvi (1982) has shown that the unmyelinated fibers or their endings in the pulp react to unnaturally high temperatures. Specific myelinated fibers react only when liquid movement

Table 1: Surface temperature of the oral soft tissue (^{o}C).

Reference	Ng (1978)		Mukherjee (1978)		Budilyna (1970)		Brill (1978)		Bergström (1971)		Teseler (1982)	
Method*	Ti		Ti		Tc		Tc		Tc/Ti		Ti	
mouth closed/open	?		?		open		open		closed		open	
Localization#	Min	Max	Min	Max	Min	Max	Min	Max	Min	Max	Min	Max
GINGIVAL SULCI												
Anterior u	31.5	33.8	32.9	35.0								
l	33.1	34.6	33.4	35.1								
Posterior u	33.4	35.2	33.5	35.5								
l	33.8	35.3	34.1	35.4								
GINGIVA												
Anterior u							33.2	35.1			34.3	34.3
l											34.8	35.1
Posterior u							34.5	35.5			34.2	35.0
l											34.9	35.4
ALVEOLAR MUCOSA												
Anterior u					34.4	34.7	33.7	35.7	35.7	36.1		
l					35.1	35.4			35.4	36.4		
Posterior u					35.8	36.2	35.1	36.1	35.7	36.5		
l					35.7	36.0			36.0	36.8		
PALATE												
Anterior					32.1	32.1			35.5	36.3		
Posterior					33.1	34.9			34.7	35.7		

* Ti = thermistor; Tc = thermocouple

u = upper; l = lower

is possible after uncovering the dentinal tubules.

SURFACE TEMPERATURE

The local surface temperatures of the oral tissues differ from site to site depending on several factors within the region (Budylina 1970) like metabolic activity, local blood flow and environmental temperature.

Methods for recording local temperatures at the surface of hard and soft dental tissues are mainly based on thermo-electrical techniques. Electric potentials arise in a thermosensitive bimetal as used in thermocouples and in thermistors. A thermistor is a semi-conductor which shows an alinear increase of voltage with increasing temperature. The thermocouple (Brown 1966; Budylina 1970; Bergström 1971; Brill 1978; Maeda 1979) as well as the thermistor (Bergström 1971; Banes 1978; NG 1978; Mukherjee 1978; Arnold 1979; Teseler 1982) have been used frequently for temperature measurements in the oral cavity.

Under certain circumstances infrared thermography was used for detection of the temperature distribution at the surface of teeth (Quenneville 1976). Crandell (1966) used an infrared thermometer for measuring the surface temperature by recording the quantity of heat at a given distance of the object. Howell (1970) and Quenneville (1976) used certain liquid crystals for thermography of anterior teeth. The temperature corresponds with the color of the crystals being calibrated by means of a thermocouple.

Registration of the local surface temperature is strongly affected by the environment. The experimental conditions as described in literature show a wide variation, i.e., variation in room temperature, the use of rubberdam, measurements with closed or open mouth. In reviewing the results of several investigations, a differentiation is made into two categories: A) soft dental tissues and B) teeth.

Soft dental tissues

Most of the temperature measurements at the surface of soft tissues were used for possible indications of inflammatory processes. The minimum and maximum surface temperatures of the soft tissues are given in Table 1. The following conclusions can be drawn from the literature (Budylina 1970; Bergström 1971; Brill 1978, Mukherjee 1978; NG 1978; Teseler 1982):

- the temperature is increasing distally;
- in transverse direction a symmetry in temperature does exist (Bergström 1971);
- the sulcular temperature in the lower jaw is higher ($0.7 \pm 0.2^{o}C$) than in the upper jaw (Mukherjee 1978);
- the buccal and lingual sulcular temperatures are equal (Mukherjee 1978);
- in comparison with the lower jaw the surface temperature of the gingiva of the upper jaw is 0.4 to $0.9^{o}C$ higher (Teseler 1982);
- the temperature of the alveolar mucosa is on average $0.7^{o}C$ higher than the attached gingiva (Budylina 1970; Brill 1978).

Teeth

Mainly the anterior teeth have been used for temperature measurements of tooth surfaces. Both vital and non-vital teeth were involved (Crandell 1966; Howell 1970; Budylina 1970; Banes 1978). The results can be summarized as follows:

- each tooth has a temperature gradient, decreasing in incisal direction (Brown 1966);
- the surface temperature of vital teeth is not systematically higher as

Table 2: Tooth surface temperature (°C).

Localization*		Anterior Min	Anterior Max	Posterior Min	Posterior Max	Method	Test condition	Reference
Buccal		28.8	32.4			thermocouple	rubberdam	Brown (1966)
Palatal		29.6	32.6					
Buccal		30.5	31.3			infrared thermometer	open mouth	Crandell (1966)
Buccal		27.4	28.8	32.1	34.7	thermocouple	open mouth	Budylina (1970)
Palatal			30.3				nose breathing	
Buccal	u	33.7	35.1			thermography	rubberdam	Howell (1970)
	l	33.8	33.8					
Buccal	u	29.9	31.0	31.5	34.9	thermistor	open mouth	Banes (1978)
	l	30.9	32.4	33.1	34.2			

* u = upper; l = lower

compared to non-vital teeth (Brown 1966; Crandell 1966; Howell 1970);
- at the palatal side of the teeth, the temperature is about 0.4°C higher than at the labial side;
- the temperature of teeth in the lower jaw is higher than of teeth in the upper jaw;
- the surface temperature of teeth increases distally.

Only the extreme values of the surface temperatures of teeth are given in Table 2.

DISCUSSION

Temperature measurements at the surface of the tissues in the oral cavity have been carried out mostly with a thermistor or a thermocouple. The reaction of a thermistor to temperature changes is slow and non-linear. Besides, the recordings are inconsistent when the thermistor is not immersed the full 3 mm into the medium (Mukherjee 1978). Therefore, a thermistor is an unreliable instrument for measuring tissue surface temperatures. Crandell (1966) used the infrared thermometer for measuring tooth surface temperatures. Recordings were made from a distance leading to interference of the environmental radiation. This instrument is unreliable as well since it is almost impossible to control the extra- and intraoral environment. Thermography as used by Howell (1970) and Quenneville (1976) results in color changes, corresponding with a temperature range. This method as well as the infrared thermography as used by Quenneville (1976) gives an indication about the level of surface temperature but cannot be used for more accurate results.

The reaction of a thermocouple to temperature changes is fast and accurate. The thermosensitive junction is tiny and easy to use in the oral cavity. Until now the thermocouple seems to be the most accurate instrument for measuring surface temperatures. However, errors can be easily made and the easiness of use could be deceptive.

The diversity of experimental conditions between various studies is great. Maeda (1979) analysed the influence of age of 40 experimental subjects and their corresponding core temperature in relation to the oral temperature. The core temperature was measured sublingually. The expected correlation between palatal and sublingual measurements was not confirmed. The factor age can therefore be considered to be of secondary importance.

The surface temperatures of teeth and surrounding tissues can be affected by the vitality of the pulp (Banes 1978; Teseler 1982). On the other hand, both Brown (1966) and Crandell (1966) have assumed a passive conduction of heat from the periodontal tissue rather than from the pulp itself. The activity of the pulp tissue will be a factor of importance and should be taken into account in measuring tooth temperatures as long as its influence on the temperature is a point of discussion.

With regard to experimental conditions, the use of the rubberdam (Brown 1966; Howell 1970; Banes 1978) can be criticized since, this is not representing a physiological situation. In Table 1 differences can be noticed between the values found by Bergström (1971) and those from Budylina (1970) and Brill (1978; mean 1.2°C higher). The difference has to be attributed to the experimental conditions: Budylina (1970) and Brill (1978) measured with the mouth opened in contrast to Bergström (1971). In this context a closed mouth can be regarded as the physiological situation. Boehm (1972) investigated the influence of open-mouth breathing on the thermal environment. Specifically, the ambient temperature, humidity, and the dryness of the cavity did affect the environment.

Table 3: Approximate surface temperature in the oral cavity (°C).

Localization	Anterior		Posterior		Reference
	Min	Max	Min	Max	
Teeth	30.0	33.8	33.3	35.9	calculated
Gingival Sulci	33.4	35.1	34.1	35.5	Mukherjee (1978)
Gingiva	34.7	35.7	35.0	36.1	calculated
Alveolar Mucosa	35.4	36.4	35.7	36.8	Bergström (1971)
Palate	35.5	36.3	34.7	35.7	Bergström (1971)

The results so far can be summarized as follows: the experimental conditions were not uniform and are frequently described incompletely. The diversity in experimental conditions and in measuring sites as well as in the number of measurements does not allow for comparison of data. In view of these limitations, an approximation of surface temperatures in the oral cavity as derived from literature is reflected in Table 3. The surface temperature of teeth is the mean temperature of the results found by Brown (1966) and Budylina (1970) plus 1.2°C because of the test conditions. For measuring the sulcular temperature, a thermistor has to be inserted into the sulcus at least 3.0 mm (Mukherjee 1978). NG (1978) however did not describe the depth of insertion. The temperatures of the gingival sulci are always lower than the real temperature values, caused by an ischemia during the testing period of about 30.0 sec per measurement (Mukherjee 1978). Mukherjee (1978) has noticed this phenomenon, but did not adapt the results. Therefore only the values of the maximum temperature are recorded in Table 3. The surface temperature of the gingiva cannot be found directly from literature. However, both Brill (1978) and Budylina (1970) have found a temperature difference of 0.7°C between the mucosa and the gingiva. The gingival temperature was calculated by subtracting 0.7°C from the mucosal temperature.

The above results will be used for a further study of theoretical modeling as described by de Vree (1983).

CONCLUSIONS

The following conclusions can be drawn:

- The system of heat regulation within teeth is still a matter of dispute.
- The thermocouple seems to be the easiest instrument to use for the surface temperature measurements.
- The temperatures of soft tissues and anterior teeth can be greatly modified through the environmental conditions.
- The mean surface temperature of the soft tissues in the oral cavity varies between 32^{o}C (anterior) and 37^{o}C (posterior).
- The mean surface temperature of anterior teeth is 30^{o}C and 35^{o}C for posterior teeth.

2.2 THERMAL TRAUMA TO TEETH*

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ABSTRACT - The literature is reviewed with respect to several thermal loading situations and their influences on the temperature field in teeth.

Breathing, food consumption, and dental treatment are processes disturbing the thermal balance in the oral cavity. The tooth temperature will be affected. These thermal changes result in a physical after-effect on enamel, dentin, and restorative materials, as well as in a histological change of the pulp tissue. Postoperative discomfort is another after-effect reported as result of dental treatment. Recommendations concerning cooling technique and the use of a cement base are given to avoid this discomfort.

In the literature reviewed, attention is mainly focused on temperature within teeth. However, the underlying thermal processes are important in the reported after-effects. It is concluded that more attention should be paid to this phenomenon in future research.

INTRODUCTION

To support a healthy constitution of the human body several physiological processes are balanced. A disturbance of equilibrium, mainly bacterial in nature, may lead to pathological changes. Nevertheless a thermal etiology may be possible as well. This review deals with thermal aspects of the oral cavity. Particularly, attention is drawn to thermal processes within teeth under biological conditions and in relation to dental treatment.

Thermocycling resulting from the intake of food and due to dental treatment might be responsible for the after-effect of thermal changes in human teeth. In addition to thermal postoperative discomfort (Piperno 1982; Silvestri 1977), it is known that enamel, dentin, pulp tissue, and restorations can be damaged by exposure to thermal fluctuations and extreme thermal shock, even resulting in necrosis of the pulp tissue (Langeland 1960; Zach 1965; Nyborg 1968; Guzman 1969; Roydhouse 1970; Lloyd 1978a).

To gain insight in the thermal aspects of the oral cavity the literature was reviewed with respect to several thermal loading conditions and their influence on the temperature field in teeth, as well as on the after-effect within teeth and restorative materials. The results are reported and some recommendations will be given concerning the protection of soft and hard oral tissues, as well as restorative materials protective against thermal shock.

Table 1: Temperature changes (^{o}C) in teeth as a result of different heat sources

Reference	Heat source positive(p) negative(n)	Temperature rise(r) decrease(d)	Initial temperature	Loading side	Measuring site tooth (restored/unrestored)
Hensel	p 50.0-65.0	r (0.6 $\pm$ 0.2)-(3.3 $\pm$ 1.0)	36.0	buccal	PDJ (buccal side)
(1956)	n 10.0-25.0	d (0.2 $\pm$ 0.1)-(1.7 $\pm$ 0.7)	36.0		upper incisor (restored)
Braden	p 42.2	r 6.7	25.2	occlusal	dentin (3.7mm)
(1964a)					molar (unrestored)
Plant	p 50.0	r 0.3-0.9	37.0	crown	PDJ
(1974)	p 60.0	r 2.0	37.0		premolar (-)
White	p 57.8	p 19.1	26.1	buccal	pulp cavity
(1977)	n skin refrigent	d 17.1	26.1	gingival third	cuspid (unrestored)
Trowbridge	p 68.0-76.0	r 3.9 $\pm$ 1.2	34.0	occlusal	PDJ
(1980)	n 14.0-20.0	-		buccal ridge	premolar (-)
Peters	n dry ice	d (1.9 $\pm$ 0.2)-(3.3 $\pm$ 0.2)	body-temperature	buccal	-
(1981b)					-
Augsburger	n1 CO_2 snow	d a: 2.23 $\pm$ 0.07 (1)	body-temperature	buccal	PDJ
(1981)	n2 skin refrigent	1.14 $\pm$ 0.04 (2)			molar:
		0.58 $\pm$ 0.03 (3)			a) unrestored
	n3 ice	b: 4.56 $\pm$ 0.68 (1)			b) full crown
		1.45 $\pm$ 0.09 (2)			
		2.06 $\pm$ 0.23 (3)			

THERMAL LOADING CONDITIONS

Generally, teeth are very adequately insulated by lips, cheeks, and tongue when the mouth is closed. However, the thermal balance can be disturbed by breathing, consumption of food and dental treatment, and these will affect the tooth temperature.

Respiration

The air temperature in the oral cavity should equal the temperature of the body. However, oral breathing as well as nasal breathing, the ambient temperature, humidity, and dryness of the oral cavity will affect the temperature field of the environment of teeth (Boehm 1972). The temperature range in the oral cavity during extreme conditions could not be derived from the literature.

Food

The consumption of food can cause thermal fluctuation (thermocycling) in the oral cavity. The temperature range of food may vary between 7°C and 75°C resulting in a temperature range from 5 to 48°C in the environment of teeth (Rothwell 1959; von Gräf 1960). Liquids at extremely low or high temperatures will result in oral discomfort. However, from the studies of Peterson (1966), Plant (1974), and Mesu (1983) it can be concluded that, under normal drinking conditions, a temperature range for liquids between 5°C and 60°C is acceptable, resulting in a temperature range from 15 to 50°C in the environment of teeth. However, some discomfort has been reported (Plant 1974) when the contact time between the liquid and the teeth is longer than 15 seconds. This temperature range is also used for experimental research on material properties (Brown 1978).

Dental treatment

Dental treatment is mostly accompanied by heat production. A lack of data in the literature concerning the temperature at the <u>surface</u> of dental tissue during restorative procedures renders it impossible to cite a temperature range. In the literature, interest is mainly focused on the temperature rise <u>in</u> teeth - during cavity preparation, setting of restorative material, and polishing. These procedures and their thermal influence will be discussed separately below.

TEMPERATURE IN TEETH

Under conditions of normal nasal or oral <u>respiration</u>, Harvey (1943) recorded temperatures <u>in</u> a restored buccal cavity of a lower molar tooth of 35.6°C and 23.3°C, respectively. Since no contradictory information is available, the temperatures recorded by Harvey (1943) might be considered as a physiological normal overall temperature range in a tooth.

The temperature <u>in</u> teeth during the consumption of <u>food</u> has been recorded by von Gräf (1960). A thermoprobe was completely imbedded in the enamel at 0.5 mm from the outer surface. The cooled food (-7°C and 5°C) caused a minimum temperature in the enamel of 16°C. Consumption of hot food (75°C) resulted in a maximum temperature in the enamel of 45°C.

Apart from these thermal loading conditions, several <u>in vivo</u> and <u>in vitro</u> experiments are given in the literature using different thermal load situations. Lisanti (1952) recorded the pulp temperature of <u>vital</u> teeth in dogs. A thermode placed in a buccal cavity had a temperature range of 51.6 to 315.6°C. The heat source was applied for 5 to 60 sec, resulting in a temperature rise at the pulpo-dentinal junction (PDJ) of 1.9 to 24.7°C and 7.5 to 50.8°C, respectively.

Results of in vitro experiments with human teeth.

In Table 1, the increase and decrease of the intratooth temperature caused by a heat source have been summarized. Only White (1977) recorded a large rise and fall in temperature in the pulp cavity, of 19.1°C and 17.7°C, respectively. In Table 1, none of the experimental conditions are similar; therefore, no definite conclusion can be drawn from these data.

To gain an idea of the degree of temperature rise and heat generated during restorative procedures, the phases of preparation and restoration are described separately. During recent decades the techniques of dental procedures have changed; therefore, the literature review is confined to the last 2 decades.

Preparation phase

During preparation, non-efficient energy is transformed into heat, but the amount of heat transmitted to the tooth depends partly on: A) the type of bur; B) the technique of cooling; and C) the speed of the rotary instruments.

The choice of bur, cooling device, and the speed depends on the operative technique. With the intracoronal operative technique, three steps can be distinguished: A) preparation of the cavity; B) excavation; and C) creation of more retention - like pinholes or self-threading pins. The last technique causes a temperature rise in the pulp of 1.7 to 5.5°C, depending on the depth of the canal (Cooley 1980). In general, no coolant is recommended for either excavation or drilling pinholes.

A coolant, however, is essential in cavity preparation, shielding the dental tissue from thermal damage. Carson (1979) has compared the water-air coolant and the air coolant by means of thermography. At a depth of 1.5 mm into the dentin, no significant differences in temperature were found between the two cooling methods. A special means of cooling is the "washed-field technique" as described by Zach (1962). The underlying principle is that the coolant flows over the tooth for 5 sec before the bur is allowed to contact the tooth. Using this technique with water-air coolant at 230,000 rpm, the intrapulpal temperature was found not to rise above the initial temperature.

In the various studies concerning the increase of intrapulpal temperature as a result of cavity preparation, a carbide or a diamond bur has frequently been used. There is no difference in heat production using several sorts of carbide burs, whereas a significant difference in heat production has been reported between the use of a diamond and a carbide bur (Consani 1976). The diamond stone showed a significantly higher heat production at the 1% level. Based on these experiments, Table 2 is composed of data from the literature related to the intracoronal preparation technique.

Comparing the extracoronal with the intracoronal preparation technique, the temperature increase appears to be identical. Preparing a crown at a speed of 160,000 rpm (technique of cooling unknown), a temperature increase of 3.2°C was recorded, but at 290,000 rpm (air-water coolant) a temperature decrease was found of 2.7°C (Schuchard 1961). However, when cutting a slice with a cylindrical bur, an extreme temperature increase has been recorded of 13.6°C (Eifinger 1979). According to these investigators, this extreme temperature rise can be attributed to insufficient irrigation of the operative region.

In summary, it can be stated that, although the type of bur, cooling

Table 2: Temperature changes within the pulp cavity during intracoronal preparation with a coolant

Reference	Type of bur (in vivo/in vitro) carbide	diamond	Revolutions per minute (coolant*)	Temperature (^{o}C) rise	decrease
Schuchard (1961)	in vivo		160,000 (?)	2.2-3.7	
			290,000 (a-w)		4.8
Zach (1962)	in vivo		240,000 (a-w)	2.2	
Bhaskar (1965)	in vivo		250,000 (w)		8.1
Consani (1976)	in vitro		250,000 (w)		4.9-5.2
		in vitro	250,000 (w)		3.2
Lloyd (1978b)	in vitro		240,000 (w)	5.0	
Eifinger (1979)		in vitro	250,000 (w)	6.5	

* a-w = air-water; w = water

system, and rotating speed are important factors in heat generation, adequate irrigation of the operative area may be even more important. The "washed-field technique" has been shown to be an effective cooling procedure.

For the combination of extracoronal preparation and indirect casting technique, an impression is required. Using rubber impression materials such as silicone, polysulfide, and polyether, temperature rises of 1.1^{o}C, 3.4^{o}C, and 4.2^{o}C respectively have been recorded during setting (Craig 1980). These amounts of heat release will not damage the underlying tissues under normal circumstances. However, using a reversible hydrocolloid impression material, a paste is ejected around the preparation at 63 to 66^{o}C without tempering. Then a tray (46 $\pm$ 1^{o}C) is placed in the mouth. The paste sets to gel between 37^{o}C and 45^{o}C by cooling the tray. In the literature no information is found concerning the consequences of this procedure, which might result in a certain amount of damage to the tissue.

Restorative phase

Plastic restorative materials usually contain more than one component. Mixing these components might result in an exothermal reaction. Such an exothermic setting reaction takes place in restorative materials such as cement, composite, and resin.

In an in vitro study, Plant (1974) and Crisp (1978) recorded divergent temperature changes in the restoration during setting of cement, composite, and resin. Similar experiments, as carried out in vivo by Bausch (1982), revealed maximum temperatures of 36.0^{o}C, although the initial temperature of the materials was not given. The data concerning the rise in temperature during setting of restorative materials are compiled in Table 3.

The resin used for temporary restorations polymerises under an enormous release of heat. The setting of this material and the heat production involved takes place mainly in the oral cavity. Thus, the prepared tooth will be exposed to extremely high temperatures. In an in vitro study, Grajower (1979) has observed a maximum intrapulpal temperature of 74.4 ± 2.0°C during curing of a temporary acrylic resin crown by means of a copperband. However, by using a silicone impression of the unprepared tooth as a matrix, the temperature in the pulp was only 3.5 ± 0.9°C higher than the initial temperature (37.0°C).

The curing of composites also represents an exothermic reaction. It is expected that transmission of excessive heat to the pulp may occur during setting of this restorative material. Recent research by Bausch (1982) has shown that a temperature increase in the composite (3.4 to 4.1°C) resulted in a temperature decrease of the pulp of 1.0°C. However, the decrease in pulp temperature was attributed to the low initial temperature of the composite materials being stored in the refrigerator before use. Thus, it can be concluded that the reported rise in temperature of the materials (Table 3) will not result in an extreme temperature increase of the pulp tissue.

Polishing

The restoration phase is finished by polishing the restorative material. Relatively little is known of the thermal processes during polishing. Aplin (1967) has investigated the temperature change just underneath (0.40 ± 0.02 mm) a Class V metallic restoration. The thermal effect of increasing the polishing speed from 2700 to 4700 rpm resulted in a temperature increase of nearly 3.3°C, independent of the polishing technique used. In the same study, the authors observed the greatest temperature rise in using a rubber cup without any cooling or polishing paste. However, using a wet polishing paste, only a small rise in temperature has been reported. Given their test conditions, Aplin (1967) recorded a temperature range from 2.4 to 11.6°C. Christensen (1968) has studied the intrapulpal temperature during polishing (2600 rpm) of a Class V amalgam restoration for 60 sec. A temperature rise of only 0.00 ± 0.18°C and 0.6 ± 0.4°C was recorded for polishing with, respectively, a green stone using air-coolant and a hard brush with moist pumice. On the other hand, polishing with a rubber disk without a coolant generated maximum heat, increasing the intrapulpal temperature with 6.2 ± 2.5°C. It was also observed that the temperature rise continued after the polishing instruments were removed.

Recently, this heat production at the surface of restorative materials has been advocated on purpose to improve the surface layer of micro-fine composites (Bausch 1981; Davidson 1981).

Table 3: Temperature increase within restorative materials during setting

Restorative material	Temperature increase (°C)		
	Plant (1974)	Crisp (1978)	Bausch (1982)
Methylmethacrylate	4.4 ± 0.6		
Composite:			
Adaptic®	2.2 ± 0.1		
Estilux®			3.4 - 4.1
Calcium Hydroxide	2.2 ± 0.3		
Zinc Oxide and Eugenol	0.0	6.5 - 9.6	
Polycarboxylate	1.2 ± 0.1	7.6	
Zinc Phosphate	2.1 ± 0.4		

AFTER-EFFECT

In the enamel crown, and especially in the cervical third part, thermal cracks and propagation of cracks are caused by thermocycling (Lloyd 1978a; Brown 1972). Fractures in tooth structure at 1.0 to 2.0 mm from the cavity wall are also induced by dry cutting (Brown 1978).

Burn lesions might occur in areas where the bur contacts the dentin, notwithstanding the presence of a cooling device (Langeland 1960).

Amalgam, as well as composite restorations, also show fractures and propagation of microcracks as a result of thermocycling (Roydhous 1970; Williams 1983; Montes-G 1983) which might cause microleakage (Guzman 1969; Peterson 1966).

Next to the physical after-effect on enamel, dentin, and restorative materials, the histology of the pulp tissue changes (Langeland 1970). The initial state of the pulp is an important factor in the degree to which the tissue will change. In operative procedures the frictional heat can reach the pulp, damaging vessels and cells, and parts of the pulp become necrotic.

According to Zach (1965), a temperature increase in the pulp of 5.6°C causes necrosis in the pulp tissue of sound teeth in 15% of cases. However, if the state of the pulp is already irritated due to caries, a moderate amount of heat trauma can even be fatal. Local necrosis may occur initially (Nyborg 1968). In other cases, thermal irritation or thermal shock caused local necrosis of the adjacent pulp tissue. Inflammatory reactions were only observed in the presence of infectious material (Watts 1979).

These findings are related to postoperative discomfort, being one aspect of the after-effects. From the literature it is known that 38 to 50% of the patients complain of thermal postoperative discomfort, despite the presence of a calcium hydroxide base (Piperno 1982; Silvestri 1977). Piperno (1982) instituted a 6-month follow-up after treatment. After one week, 21% of the patients complained of thermal discomfort, but after 6 months all treated patients were free of complaints. The reaction of the pulp tissue and the lack of reparative dentin were positively correlated to postoperative discomfort.

This postoperative discomfort has to be prevented during the preparation stage by a treatment as atraumatic as possible and by protection of the dental tissues against negative external influences. What has not been described, however, are the external influences which resulted in thermal hypersensitivity after treatment, as suggested by Piperno (1982) and Silvestri (1977).

DISCUSSION

The temperature range at the surface of teeth under various thermal conditions and their influence on the intratooth temperature have been reviewed. However, the reported findings concerning several thermal problems are hardly comparable. For instance, the thermal loading conditions vary as to site, duration and temperature. Differences occur in measurement site, in distance between heat source and measurement point, and in the case of dental treatment in the amount of heat production. Standardized experiments to simulate thermal loading conditions in a tooth model during preparation or restoration may resolve this problem. Only Aplin (1967) used replicas of resin of the same tooth with a Class V preparation in it. Westland (1978) introduced a standard measure for grinding energy. To efficiently remove 1 unit volume of a particular

material, a certain amount of energy is necessary: the so-called "Specific Grinding Energy" (SGE).
In formula:

$$SGE = \frac{M\ \omega\ T}{Q} \qquad (\text{Joule.mm}^{-3})$$

(where M = drill-moment in Nm; ω = angle-speed in sec^{-1}; T = grinding-time in sec; Q = grinding-volume in mm^3). The SGE value also gives a non-efficient energy value, which evolves into heat. By using this SGE as a measure in a standardized experiment for preparation, a measure of non-efficient energy is given from which the expected intratooth temperature can be calculated for any situation.

In the case of setting restorative materials, the variation in cavity geometry results in different intratooth temperatures. Also the site of measurement at the restoration-dentin interface influence the temperature values. Moreover, the quality of the dentin may change with increasing age, resulting in an increase in thermal conductivity. A mixed content of dentin with reparative dentin may also change the temperature records.

The temperature changes recorded in Tables 1 and 2 are the data of in vitro as well as in vivo experiments. The various experimental conditions of the pulp content and the fixation of the tooth resulted in the differences in results. For instance, Consani (1976) did not use any pulp content, but the tooth was placed in a dental plaster, which reduced the temperature rise by 50% (Heithersay 1963). Eifinger (1979) filled the pulp cavity with a thermoplastic mass resulting in a temperature increase of 6.5°C. However, the in vivo measurement of Bhaskar (1965) shows a decrease of 8.1°C. Therefore, no correlation can be found so far between laboratory thermal studies and clinical findings.

The energy release during setting of restorative materials results in small temperature changes (Table 3), which hardly affect the intrapulpal temperature.

The intrapulpal temperature rise during polishing, as recorded by Christensen (1968) is not only the result of the distance between polishing surface and measuring site, but is also caused by the stone in which the tooth was embedded. This resulted in insulation of the model, except that the polishing site of the tooth was not covered. Since polishing can lead to a temperature rise in the pulp, it is clinically important to reduce the polishing time for reasons of prevention.

In view of the limited recuperative potential of the tissue, it is important to reduce the risk of temperature changes in the pulp tissue. Therefore, insulating restorative materials can favor this. For instance, microfine and highly filled composites are restorative materials with low thermal diffusivities, which are respectively equal to dentin and within the range of enamel and dentin (Watts 1983). However, in the posterior region, amalgam is still superior to composite, despite its high thermal diffusivity. Therefore, cement bases are used in combination with amalgam for providing thermal insulation. Cavity varnishes do not have any effect on the thermal diffusion in restored teeth (Voth 1966).

The thermal hypersensitivity of pulp tissue after dental treatment is partly due to insufficient insulation. In amalgam restored teeth, the efficiency of cement bases in providing insulation is dependent on its thickness and its thermal diffusivity (Braden 1964a). The thickness of the base is the most important parameter in so far as thermal insulation is concerned (Voth 1966). However, the modulus of elasticity limits the thickness of those cements (Farah 1983), resulting in limited thermal protection. Generally, in vivo and in vitro experiments have shown a reduction in temperature rise at the cement-dentin interface using zinc phosphate, calcium hydroxide or zinc oxide-eugenol cement (Voth 1966; Tibbetts 1976). The significance of these results for the temperature in the pulp is not known. A theoretical study by Spierings (chapter 4.1) reports a small effect in temperature change at the PDJ when using a calcium hydroxide base. Notwithstanding these results, a cement base is recommended beneath an amalgam restoration for limiting thermal trauma as well as initiating the production of reparative dentin. Theoretical studies using finite element analysis may enlarge the insight in temperature distribution in human teeth.

CONCLUSIONS AND RECOMMENDATIONS

The following conclusions can be drawn:

- In the literature, no information is found concerning thermal distribution and thermal processes in human teeth.
- The heat production during preparation is mostly dependent on the degree of irrigation of the prepared region.
- In general, heat production during setting of various restorative materials is minimal.
- The degree of heat trauma depends on the recovery of the tissue.
- Subsequent to restorative procedures, there is postoperative discomfort such as thermal hypersensitivity in 38 to 50% of the cases.
- Dental treatment might cause a rise in intrapulpal temperature through multiple factors with unknown consequences.

In order to minimize postoperative discomfort, the following recommendations can be given:

- During the preparation phase, insufficient irrigation can be prevented by using the "washed-field technique".
- Application of an insulating base as part of the restorative phase has to be emphasized.

This literature review has shown that attention is mainly focused on the temperature within teeth. To gain more insight into thermal processes within teeth, extensive thermal analyses using theoretical methods are recommended.

3.1 A SIMULATION MODEL FOR TRANSIENT THERMAL ANALYSIS OF RESTORED TEETH*

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ABSTRACT - A theoretical model was developed to analyze the influence of various cement bases on temperature distribution and heat flow in restored teeth. A physical model of a molar was developed to simulate different thermal processes by simple parameter variations.

The time-dependent temperature field was calculated using the finite element method (FEM). The values for material properties and thermal load were chosen from dental literature. The results are in good agreement with clinical experimental research as published by Trowbridge (1980). It is concluded that the model is a valid tool for further research with regard to the influence of restorative materials and cavity design on the thermal behavior of restored teeth.

* J Dent Res 1983
62: 756-759

INTRODUCTION

The thermal behavior of restored and unrestored teeth has been the subject of dental studies for many years. These activities were directed mainly toward pulpal reactions and/or thermal properties of biological and restorative materials. The introduction to dental research of advanced mathematical methods gives rise to the possibility of constructing a physical model of restored and unrestored teeth using data from previous findings (Farah 1972; Peters 1981a; Takahashi 1982). In physical modeling, the physiological reality is reduced to a system of geometrical areas containing continua that can be characterized by finite sets of well-defined parameters. In spite of the enormous amount of research done so far, there are still gaps in our knowledge prohibiting the development of a sound and conclusive physical and/or mathematical model. These gaps can be bridged, among other possibilities, by making assumptions regarding the extreme limits of missing values of parameters.

The aim of the present study is to develop a simulation model for studying a heat transport problem in a continuous system. The system comprises the coronal part of a molar, consisting of restorative materials and hard and soft dental tissues. At the boundary of this continuous system, a thermal load is prescribed, resulting in temperature changes within the system. By applying the physical laws of heat transport in continua, a mathematical model is developed consisting of a set of equations in the relevant variables. Unknown variables in this respect are the temperatures as a function of time and place.

MATERIALS AND METHODS

We chose a model of a tooth structure containing a class I occlusal restoration and with an equal thermal load at all points around the coronal part of the outer surface. For determination of the contours and dimensions of the different areas of dentin, enamel, and pulp chamber, we chose from

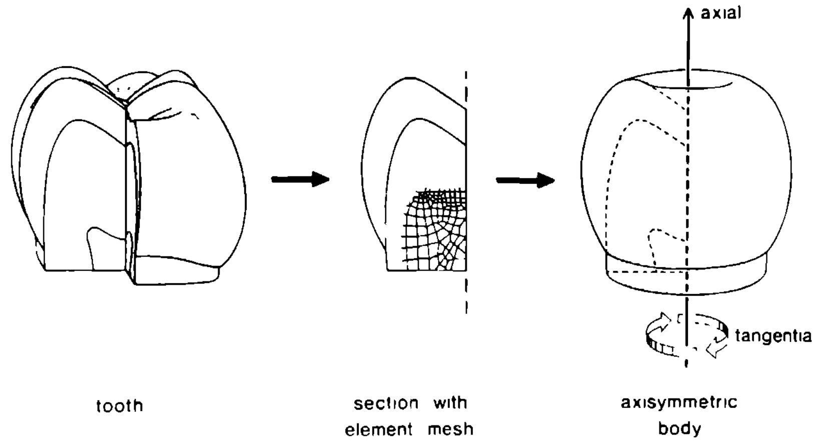


Fig 1 - The construction of an axisymmetric body from the real three-dimensional body.

values for a human mandibular first molar as reported by Kraus (1969) and as used by Peters (1981a).

The following assumptions were made:

(1) The heat transport in the tangential direction (Fig 1) is negligible. Therefore, an axisymmetric model is chosen with dimensions representative of the different subareas, Ω_i (Fig 2).

(2) The transport of heat within the tooth takes place as conduction; radiation is not significant.

(3) The temperature is continuous at the interfaces of the subareas.

(4) For the lower part, Γ_T, of the outer surface of the model (Fig 2), a constant temperature, T_0 is prescribed:

$$T = T_0 \text{ on } \Gamma_T, \text{ with } T_0 = 35.2^{o}C.$$

(5) For the remaining upper part of the outer surface of the model, $\Gamma\alpha$, the heat flux perpendicular to the outer surface, q_n, is proportional to the difference between the local unknown boundary temperature, T_b, and the local time-dependent surrounding temperature, T_∞ :

$$q_n = \alpha(T_b - T_\infty) \text{ on } \Gamma\alpha$$

The constant of proportionality, α , is called the "heat transfer coefficient" and can be a function of the coordinates and of time.

(6) At the pulpo-dentinal junction (PDJ), the reality can be expressed by two extreme conditions: (A) thermal insulation at the PDJ; and (B) heat transport in the pulp through heat conduction and convection caused by the blood circulation. Condition A produces an upper temperature limit at the PDJ, whereas condition B produces a lower limit. In case of a healthy pulp, situation A is unlikely to occur. As a safe upper limit, a third case could be chosen namely heat conduction through the pulp, but without heat transport caused by liquid flow in the pulp. Because little is known about heat transport by means of blood and lymph circulation in the pulp, we chose this third model.

(7) The different materials are assumed to have isotropic and homogeneous physical properties.

The numerical values of the physical properties - mass density (ρ_i), specific heat (c_i), and thermal conductivity (λ_i) in the subareas of dentin, enamel, amalgam, and the pulp - are taken as constants. The values

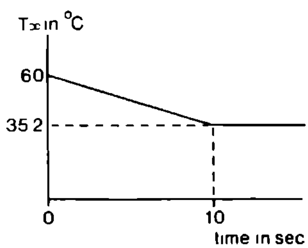
Table 1: Values of material properties used in the FEM analysis

Material	Thermal conductivity (J/m.s.°C)	Density (kg/m³)	Specific heat (J/kg.°C)	Literature reference
Enamel	9.37E-1	2.80E+3	0.71E+3	Craig (1961) Sicher (1966) Brown (1970)
Dentin	5.84E-1	1.96E+3	1.60E+3	Craig (1961) Brown (1970)
Pulp	6.30E-1	1.00E+3	4.20E+3	O'Brien (1978)
Amalgam	2.27E+2	1.05E+3	0.24E+3	Craig (1961) Vrijhoef (1981)

used in the analysis are given in Table 1. For the pulp, the values of water are taken according to O'Brien (1978).

The value of the heat transfer coefficient, α, is derived from Jacobs (1973). A rather high value of 500 J/m^{2}s°C is taken. As a loading condition, warm water is taken, with a temperature decreasing linearly in a ten-second interval from 60 to 35.2°C and afterward remaining constant (Fig 2).

For the initial conditions, a constant temperature of 35.2°C is assumed for the entire model.



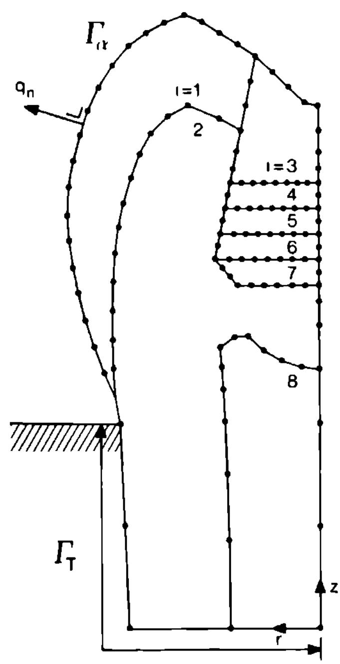


Fig 2 - The physical model showing: Boundary $\Gamma\alpha$, where the heat flux, q_n, is related to the time-dependent surrounding temperature as indicated in the graph; and boundary Γ_T, where the temperature is prescribed, and the 8 different subareas, Ω_i (i = 1,2,.....8).

Application of the heat balance law and Fourier's heat conduction law for homogeneous isotropic materials leads to the following set of partial differential equations:

$$\lambda_i \left[\frac{\partial}{\partial r}\left(r\,\frac{\partial T}{\partial r}\right) + \frac{\partial}{\partial z}\left(r\,\frac{\partial T}{\partial z}\right)\right] = \rho_i c_i r\,\frac{\partial T}{\partial t}$$

in Ω_i for $i = 1,2,\ldots..n$

The index i indicates the indexed subarea, Ω_i (Fig 2); also r = radial coordinate, z = axial coordinate, T = temperature as a function of time and coordinates r and z, n = number of subareas. The material parameters are: λ_i = thermal conductivity of subarea Ω_i; c_i = specific heat in subarea Ω_i; and ρ_i = mass density in subarea Ω_i.

From the buccolingual cross-section of the mandibular first molar an axisymmetric model was constructed as indicated in Figure 1. In this axisymmetric model the dimensions of the enamel layer, the dentinal layer, and the pulp can be easily varied.

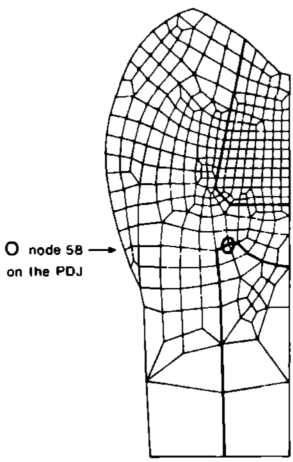


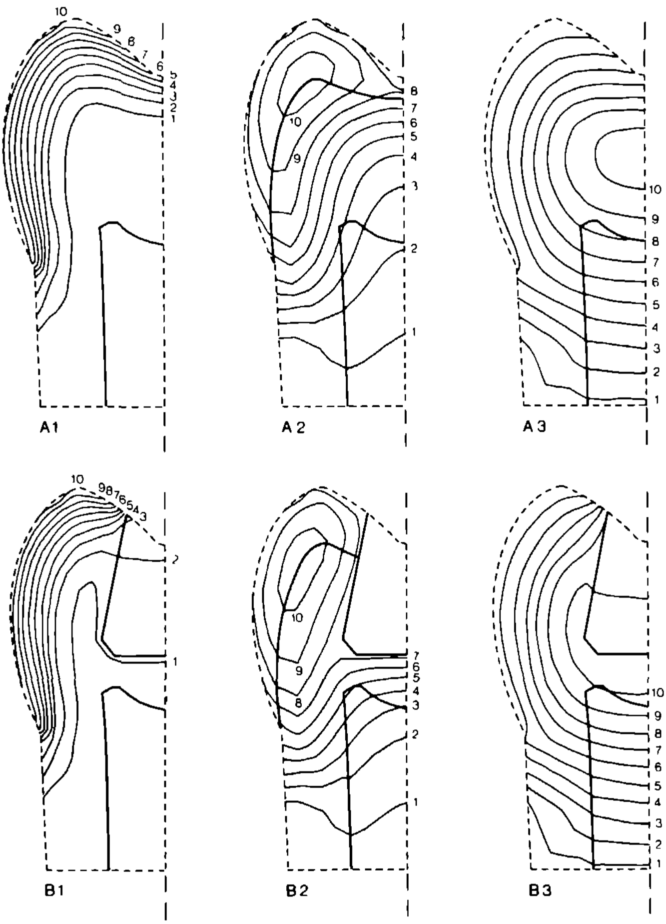
Fig 3 - The model of a restored molar divided in quadrilateral ring elements. The number 58 indicates the node on the PDJ where the highest temperatures are reached.

After the model was completed as illustrated in Figure 2, an element mesh was made using the mesh generator program TRIQUAMESH (Schoofs 1978) for two different models: (a) an unrestored tooth, and (b) a tooth containing an occlusal class I amalgam restoration (Fig 3).

A quadrilateral temperature ring element with a bilinear temperature field was chosen. Subsequently, a series of incremental calculations has been made using the Finite Element Method program (MARC®; Analysis Research Corporation, Palo Alto, CA 94306) to obtain an approximate solution to the set of partial differential equations. Therefore, the time interval of up to 30 sec after initiation of the thermal load is divided into a number of time increments. The processing was stopped after 30 sec, since the temperatures at all nodes passed their maximum value within this elapsed time.

RESULTS

The isotherms at three different times are reproduced in Figure 4. Figure 5 shows the temperature change for the PDJ at node 58 (as indicated in Fig 3). From Figure 4, we can deduce the direction of the heat flow, concurrent with the orthogonal trajectories. The density of heat flux is quantitatively proportional to the density of the isotherms. In the restored model, a quick influx of heat can be noticed at the site of the amalgam restoration, resulting in a relatively high maximum temperature at the PDJ. The maximum rise in temperature at node 58 in the PDJ is 1.8°C for the unrestored model and 2.2°C for the restored model.



Times (sec)	Isotherms (°C) 1	2	3	4	5	6	7	8	9	10
t_1 = 1.7	35.7	36.7	37.6	38.6	39.6	40.6	41.5	42.5	43.5	44.5
t_2 = 9.9	35.4	35.9	36.4	36.9	37.3	37.8	38.2	38.8	39.3	39.7
t_3 =30.0	35.3	35.5	35.7	35.9	36.0	36.3	36.5	36.7	36.9	37.1

Fig 4 - Isotherms in the unrestored (A) and the restored model (B) at three different times (t_1,t_2,t_3). A_1 refers to model A at time t_1 after initiation of the thermal load, etc. In order not to confuse the reader, the enamel-dentinal junction is represented only in the Figures A2 and B2.

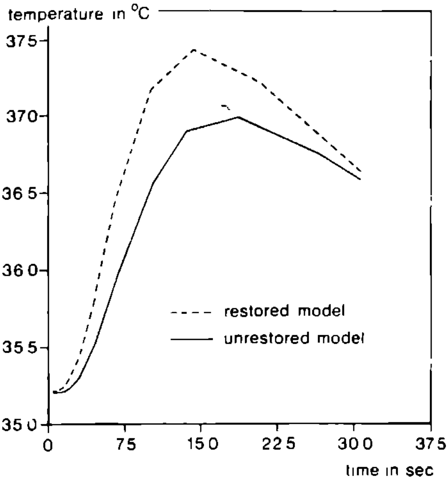


Fig 5 - The temperature as a function of time after initiation of the thermal load, at node 58 on the PDJ where the highest temperatures are reached.

DISCUSSION

Given our present-day knowledge regarding material properties, heat transfer at the outer surface, and heat transport on the PDJ, a physical mathematical model of a molar was constructed that will be used to analyze the insulating properties of cement bases. In the clinical experimental study of Trowbridge (1980) an average rise in temperature of $3.9 \pm 1.2^{o}C$ is measured at the PDJ. The thermal load in that experimental study consisted of heated gutta-percha (from 68 to $76^{o}C$) applied midway between the buccal cusp and the center of the occlusal surface. The initial temperature was $34^{o}C$. Trowbridge (1980) found a time lapse between initial loading and initial temperature change at the PDJ of 3.68 ± 1.15 sec. This is in reasonable conformity with the calculated time lapse in the unrestored model (Fig 5). The simulated thermal load lies in the range of "possibly occurring" thermal loadings and is rather high considering the chosen value for α of 500 J/m^2s^oC. As calculated in our model, the rise in temperature at the PDJ induces no severe pulpal trauma, as far as can be derived from the weak criteria given in the literature (Zach 1965). More experimental work is needed to substantiate these criteria. When a heat transport is assumed by liquid circulation in the pulp chamber, the calculated temperatures in the pulp chamber will be lower than those found in the present study.

CONCLUSIONS

The described model proved to be a reliable and valuable tool in fundamental studies of the influence of dental restorations on thermal behavior of restored teeth. This model will be used for our further studies on the influence of dental bases on restored teeth.

4.1 THE INFLUENCE OF RESTORATIVE DENTAL MATERIALS ON HEAT TRANSMISSION IN HUMAN TEETH*

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ABSTRACT - Using the finite element method, we analyzed the temperature field and heat flow patterns in an axisymmetric tooth model. The models of an unrestored tooth and three teeth restored with different restorative materials were evaluated comparatively.

The insulating ability of a calcium hydroxide cement base ($Ca(OH)_2$) is low, which is inherent to its insufficient thermal and poor mechanical properties. In the given conditions, the $Ca(OH)_2$- base reduces the temperature increase at the cement-dentin interface by 12.5% with respect to a restoration of amalgam only. By using a double base ($Ca(OH)_2$ + polymer-modified zinc oxide-eugenol), the reduction is 21.9%. Compared to a sound tooth model, the presence of a double base in the restored tooth caused a temperature increase of only 0.1°C at the pulpo-dentinal junction.

The thermal conductivity coefficient (λ) of amalgam does not have any influence on the results of the calculations. The heat transfer coefficient (α) turned out to be an essential parameter in this mathematical model.

* J Dent Res 1984 63: 1096-1100

INTRODUCTION

Restorative procedures are often injurious to dental tissues. Traumatic factors, mostly chemical or thermal in nature, may produce postoperative complaints. Hypersensitivity to cold within 24 hours after dental treatment has been reported to be experienced by 50% of the patients, whereas discomfort resulting from hot stimuli has been reported by 19% of the patients (Silvestri 1977). Cavity preparation and restoration may result in thermal loading of the tooth with subsequent irritation of the pulp tissue, resulting in hyperaemia, pulpitis, or even non-vitality. The relation between the temperature and the sensory response has been widely investigated (Hensel 1956; Zach 1965; Silvestri 1977; Trowbridge 1980; Jyväsjärvi 1982). Thermal analysis of human teeth, however, has not been frequently reported. When dental structures are substituted by restorative materials, the thermal diffusivity of the tooth will change. The use of a cement base can also affect this process.

Calcium hydroxide ($Ca(OH)_2$) and polymer-modified zinc oxide-eugenol (ZOE) are considered to be cement bases with good insulating properties. Thermal analysis of restored teeth has been used to study the influence of these bases on the temperature field.

Thermal analysis can be carried out by calculation of the temperature field in a theoretical tooth model by means of a numerical mathematical method, the Finite Element Method (FEM). In the last decade, this mathematical method has received increasing notice in dental biomechanical studies. The method has been used for two-dimensional as well as axisymmetric tooth models, analyzing the mechanical stress (Farah 1972; Peters 1981a), the thermal stress (Lloyd 1978a), or the thermal conductivity (Takahashi 1982).

The present study comprises a comparative investigation into changes of the temperature distribution in human teeth restored in various ways. Also, some parameters were examined for their effects on the temperature distribution in the tooth models.

MATERIALS AND METHODS

For thermal analysis, an axisymmetric tooth model was used, as developed by de Vree (chapter 3). The calculations were carried out according to the FEM. The geometry of the axisymmetric tooth model was defined by a bucco-lingual cross-section of a mandibular molar tooth. An element mesh (Fig 1A) was made using quadrilateral ring elements with a bilinear temperature field.

Thermal analyses have been carried out for 4 models (Fig 2): (A) an unrestored molar; (B) a molar with an occlusal cavity (depth 3.6 mm) restored with amalgam; (C) the same as B but restored with an amalgam restoration and one base, i.e., $Ca(OH)_2$; (D) the same as B but restored with an amalgam restoration and two bases, i.e., $Ca(OH)_2$ and ZOE. The thickness of the bases was 0.5 mm for the $Ca(OH)_2$ and 1.5 mm for ZOE. The geometry of the occlusal cavity was similar for the restored models B, C, and D.

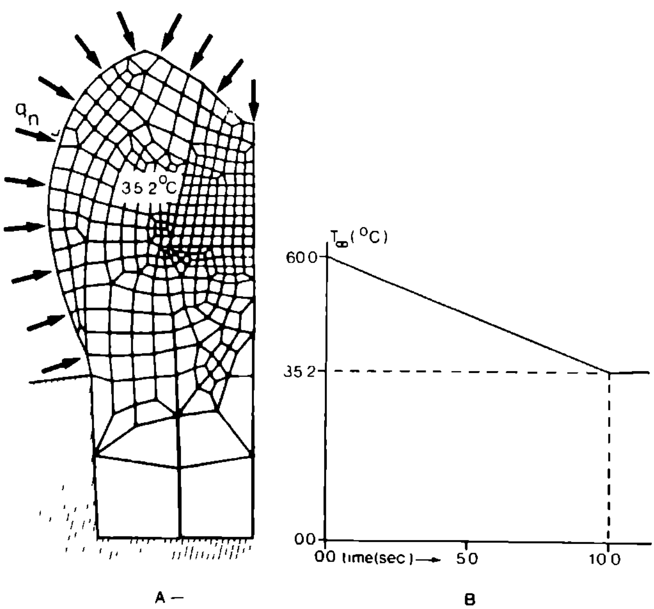
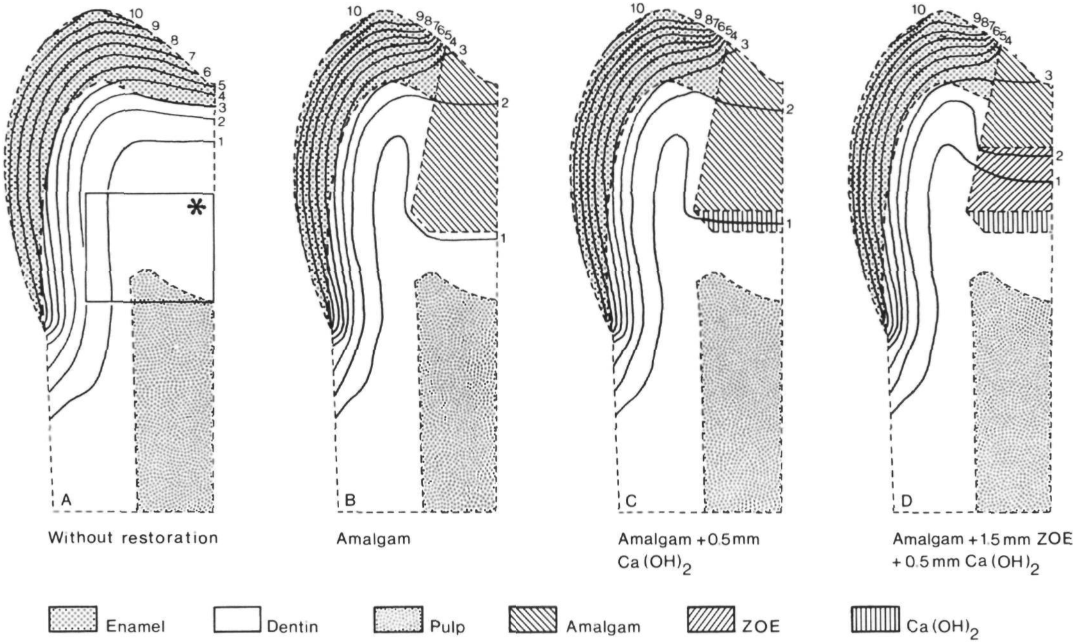


Fig. 1 - (A) Cross-section of the axisymmetric model, including the element mesh. q_n is the heat flux perpendicular to the surface. The shaded boundary at the subgingival part of the root has a prescribed temperature of 35.2°C. (B) The temperature of the surrounding liquid (T_∞) at the upper coronal part during application, as a function of time.



Isotherm	1	2	3	4	5	6	7	8	9	10
Temp. (^{0}C)	35.7	36.7	37.6	38.6	39.6	40.5	41.5	42.5	43.4	44.4

Fig 2 - The isotherms and their corresponding temperatures in the various models A, B, C, and D at 1.7 sec after initiation of the heat flux. * The restricted area shown in Figure 4.

Table 1: Material properties.

Material	Thermal conductivity (J/m.s°C)	Density (Kg/m^3)	Specific heat (J/kg°C)
Enamel(1,2,3)	$9.37.10^{-1}$	$2.80.10^3$	$0.71.10^3$
Dentin(1,3)	$5.84.10^{-1}$	$1.96.10^3$	$1.60.10^3$
Pulp (4)	$6.72.10^{-1}$	$1.00.10^3$	$4.20.10^3$
$Ca(OH)_2$ (5)	$4.66.10^{-1}$	$1.88.10^3$	$0.88.10^3$
ZOE (*,6)	$4.75.10^{-1}$	$2.29.10^3$	$0.75.10^3$
Amalgam (1,7)	$226.80.10^{-1}$	$10.50.10^3$	$0.24.10^3$

(1) Craig (1961); (2) Sicher (1966); (3) Brown (1970); (4) O'Brien (1978); (5) Brady (1974); (6) Civjan (1972); (7) Vrijhoef (1981);
* Polymer-modified zinc oxide-eugenol

For completion of the mathematical model, the following assumptions were made:

- Radiation is considered to be of no importance;
- The heat transport in a tangential direction is negligible;
- The temperature at the internal interface of the dental tissues and the restorative materials is continuous;
- The material properties (Table 1) are independent of temperature, direction, and location;
- The thermal properties of the pulp are like those of the soft tissues; thus, the material properties are equal to those of water;
- The initial temperature of the model (T_o) is 35.2°C (Fig 1A). The oral environment has an ambient temperature which will not exceed 37.0°C. The temperature of a lower molar tooth will be higher than the local surface temperature of 34.7°C (Budylina 1970). Therefore an initial temperature of 35.2°C has been assumed.
- The heat input at the coronal part of the outer surface of the model follows the formula:

  $$q_n = \alpha(T_b - T_\infty)$$

  (q_n = heat flux perpendicular to surface (Fig 1A); α = heat transfer coefficient; T_b = local boundary temperature; T_∞ = environmental temperature).
- The temperature is prescribed at the subgingival cervical part of the root: $T = T_o$ (Fig 1A). In other words, the local temperature there remains at a constant level.

The thermal load has been simulated by application of a warm liquid to the coronal part of the outer surface. The temperature of this liquid was assumed to decline linearly from 60.0 to 35.2°C in 10 sec, remaining constant during the rest of the time (Fig 1B). The heat transfer coefficient of the environment was assumed to be equal to the value of a liquid (applesauce) in steady-state, i.e., $\alpha_1 = 5.10^2 J/m^2s°C$ (Jacobs 1973).

Only limited data could be derived from the literature in regard to the heat transfer coefficient α and the thermal diffusivity of amalgam. The influence of these parameters has been studied when their values were varied. For this purpose, the influence of the thermal conductivity coefficient λ has been analyzed, comparing the temperature distribution in model B after increasing the value of the parameter λ (marked with λ_{100}) by one quarter. The new value of the parameter was marked with λ_{125}. Analysis of the effect of variation in α was done by using two different values for α: α_1 and α_{100} Value $\alpha_1 = 5.10^2 J/m^2s°C$ has been derived from literature (Jacobs 1973). The extreme value α_{100} (= $100.\alpha_1$) simulates the upper limit for the boundary conditions. A comparison was made between the results of the models B and C, since they are frequently used in the clinical situation.

RESULTS

The data resulting from the calculations can be illustrated with graphics and plots. In Figure 2, the temperature field after 1.7 sec is shown for the various models. Throughout the models, the isotherms - lines connecting points of equal temperature - are plotted. Model A shows an evenly distributed pattern of isotherms in contrast to patterns of the restored models. The temperature level for isotherm 1 is 35.7°C; the level of isotherm 10 is 44.4°C. The presence of a restoration changes the direction of the heat flow, which is perpendicular to the isotherms.

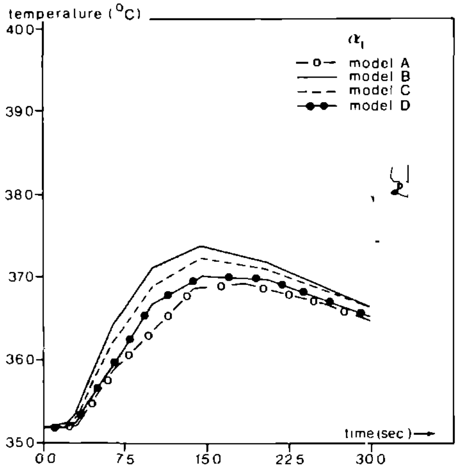


Fig 3 - The temperature change for the heat transfer coefficient α_1 at node P in the models A, B, C, and D during a period of 30 sec.

Figure 3 shows the correlation between the time and temperature at node P at the border of the pulpo-detinal junction (PDJ) in a time lapse of 30 sec. Under the given conditions the maximum temperature for model A is 37.0°C in 18.5 sec. The maximum temperature for the restored models was achieved in 14.2 sec, corresponding to 37.4°C for model B, 37.3°C for model C, and 37.1°C for model D. The presence of a combination of cement bases (model D) led to a decrease of the rise in temperature of 13.6% with respect to model B (restored with amalgam only). The $Ca(OH)_2$ in model C caused a reduction of 4.5% in temperature rise. Comparison of the temperature at the amalgam-dentin interface in model B with the temperature at the cement-dentin interface in models D and C shows the same trend. The maximum temperature at these interfaces is 38.4°C, 38.0°C and 37.7°C for model B, C, and D, respectively.

The isotherms in the area between the pulp and the restoration in the restored models are shown in Figure 4 for five different points in time. The presence of cement bases changed the direction of the heat flow. At the same time, they caused a clear slow-down in the temperature increase in the dentin and the pulp tissue, especially during the first 10 seconds.

Thermal analyses with a heat transfer coefficient α_{100} were carried out for models B and C. The temperature of the outer surface increases to a

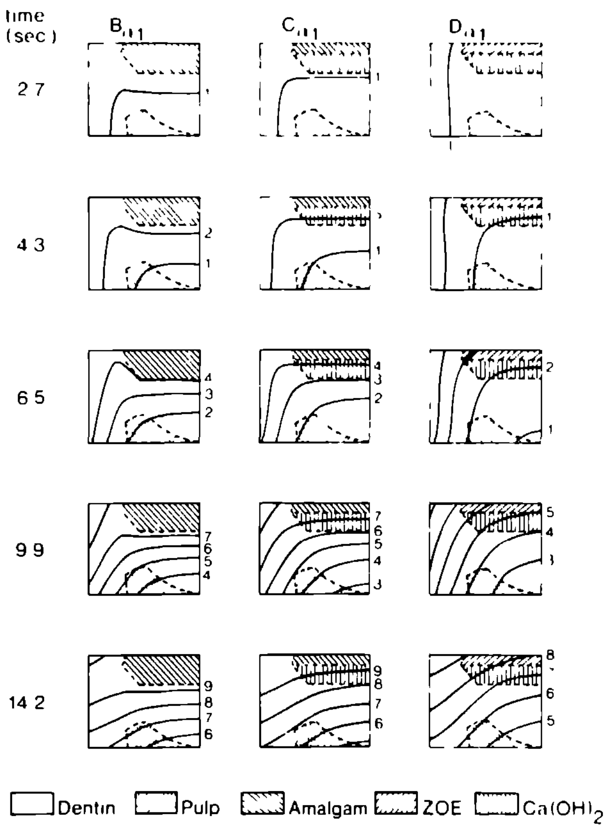


Fig 4 - The isotherms in the vicinity of the pulp and restoration for a heat transfer coefficient α_1 in the models B, C, and D at several times after initiation of the heat flux. The restricted area shown is indicated in Figure 2.

maximum at node S, the temperature curve for which is illustrated in Figure 5 for both α_1 and α_{100}. For α_1 as well as α_{100}, the results show identical temperature curves at node S for both models B and C. For α_{100} the temperature at node S has reached its maximum value within 0.3 sec, followed by a decrease. In Figure 6, the maximum rise in temperature is shown at node P for α_{100} as a function of time. Both models (B and C) have achieved the maximum rise in temperature at node P in 9.9 sec. At that time the temperature is 39.8°C in model B and 39.4°C in model C, resulting in temperature increases of 4.6°C and 4.2°C, respectively.

Thermal analyses have been carried out for the variable model parameter, thermal conductivity coefficient (λ). Input of the value of the parameters λ_{100} and λ_{125} results in identical data. Therefore the results of these calculations are not represented.

DISCUSSION

Direct experimental methods have been used frequently in the field of heat transport. Some of these studies are specifically directed to the thermal diffusivity of restored teeth (Braden 1964b; Augsburger 1981; Peters 1981b; Tibbetts 1976; Peters 1981c). Braden (1964b) showed, in an in vitro study concerning thermal diffusivity in restored teeth, that the thickness of a cement base is of more importance than its composition. However, when the thickness of a base was greater than 0.5 mm Peters (1981c) did not observe any significant reduction in the temperature increase. Using direct experimental methods, generalization of the statements derived from thermal analysis can not be justified. Particularly, the wide variations in geometry of human teeth and cavity design as well as the exact duplication of the thermal load are unsolved problems.

Numerical mathematical methods do not have these problems. However, due to lack of knowledge, some assumptions have to be made. These assumptions limit the generality of the conclusions. Nevertheless, these methods have been used progressively in thermal studies in dentistry. Lloyd (1978a) used the FEM to study the thermal stresses in an axisymmetric tooth model. Takahashi (1982) analyzed the intrapulpal thermal changes of gold-restored teeth. However, this study was restricted to a two-dimensional model in which the thermal analysis was considered to be a steady-state problem.

The FEM used is a method of approximation, reducing the physical reality of a human tooth under thermal load into a mathematical model. The results of these thermal analyses enlarge our insight into the thermal processes in human teeth. This method is also particularly suitable for analysis of the effects of variations in the parameters.

In the tooth model, the simulated load results in a transitional area in the vicinity of the cervical part of the model. This phenomenon is inherent in the modeling chosen. Obtaining more accurate results in this part of the model is possible by refining the model in this area. However, between the respective models of Figure 2, hardly any difference can be observed in the pattern of isotherms in the cervical part of the root. This is in contrast to the patterns of isotherms in the crowns of these models, which show a wide variation. Thus, this situation was deemed to be a good approximation of the physiological reality.

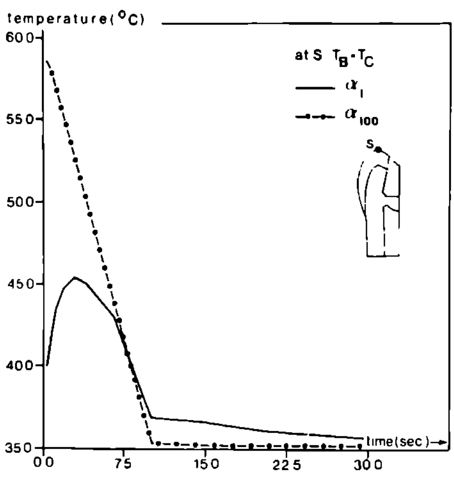


Fig 5 - The temperature change for the heat transfer coefficients α_1 as well as α_{100} at point S in models B and C. The temperatures at S in model B (T_B) are equal to those of model C (T_C).

In a comparative clinical situation, the location of node P is most vulnerable and is therefore used mostly for temperature measurements in the pulp (Hensel 1956; Trowbridge 1980). The calculations point out that both the temperature increase and the rate of temperature rise are maximal at node P on the PDJ.

The time lapse between the application of the heat source and the first response of a patient has been reported to be about 1 sec (Hensel 1956). Trowbridge (1980) has stated that the first thermal sensory response was indicated before the initial temperature change at the PDJ occurred, and these seemed to be independent of each other. This statement can be disputed. The time of initial temperature change at the PDJ observed in that study was 3.68 $\pm$ 1.15 sec. This theoretical model shows already an initial temperature change at node P within this time lapse (Fig 3). However, such a temperature increase could be undetectable in experimental methods.

The rate of temperature rise is reflected in the steepness of a temperature curve. As shown in Figures 3 and 6, the steepness depends on the presence of a restoration with or without any cement base. Therefore, it can be disputed that not only the temperature increase but also the rate of temperature rise might be of importance in the reaction mechanisms of the pulp to thermal stimuli. Accordingly, the assertion by Jyväsjärvi (1982) that the insulated nerve fibres or their endings in the pulp react directly on thermal stimulation supports this assumption.

For α_1, the temperature in model B at node P (Fig 3) has reached a maximum increase of 2.24°C. According to Zach (1965), a temperature increase of 5.6°C in the pulp of a sound tooth caused a necrotic pulp in 15% of the cases. In a restored tooth, the pulp often has been irritated by the original carious lesion and by the subsequent dental treatment. A thermal load as simulated in this study might produce pulpal injury. In view of the maximum increase of the temperature, the results make clear that the postoperative protection of the dentin against thermal stimuli is minimal. The maximum reduction of the temperature increase achieved is 0.4°C. The investigation of Silvestri (1977) supports this conclusion.

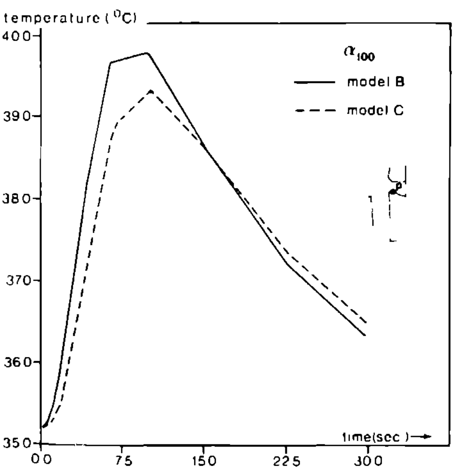


Fig 6 - The temperature change for a heat transfer coefficient α_{100} at node P in models B and C during a period of 30 sec.

Also, the maximum temperature at the restoration-dentin interface has been analyzed to achieve insight into the insulation efficiency of the cement bases. Comparing the maximum interface temperatures it must be realized that the temperatures found are directly related to the modeling chosen. The temperature at the amalgam-cement interface in models C and D is higher than at the amalgam-dentin interface in model B. Thus, with respect to model B, a cement base of $Ca(OH)_2$ reduces the temperature increase by 12.5% at the cement-dentin interface. By using an extra cement base of ZOE (model D) the reduction in temperature increase is 21.9%.

From Figures 1 and 5 it can be derived that, for α_{100}, the boundary temperature is equal to the surrounding temperature, T. So, the quantity of heat flow is maximal for α_{100}. The efficiency of the insulation of the $Ca(OH)_2$-base (model C) is 8.6% with respect to model B under the extreme thermal loading conditions (Fig 6). For α_1 this is only 4.5% in model C. Comparing Figure 6 with Figure 3 it can be noticed that the maximum temperature at node P in Figure 6 was higher and was reached earlier (4.3 sec). Lack of sufficient information about the real value of α and the demonstrated importance of this parameter when analyzing thermal problems will give direction to future experimental research.

4.2 THE INFLUENCE OF CAVITY GEOMETRY ON HEAT TRANSMISSION IN RESTORED TEETH*

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ABSTRACT - The temperature distribution and the heat flow pattern in axisymmetric tooth models were studied using finite element analysis. Variation in cavity geometry and restorative material (amalgam with and without a base) were evaluated comparatively.

In the vicinity of the pulp and the restoration, the direction of the heat flow is independent of cavity geometry. However, with respect to a standard occlusal restoration, an amalgam buildup showed an increase in both the maximum temperature and in the rate of temperature rise at the pulpo-dentinal junction depending on the presence of a cement base. Besides the cavity geometry, the temperature field depends also on the properties of the restorative materials used.

Under the given conditions, the application of a cement base cannot be regarded as efficacious for protection of dentin against thermal stimuli in the postoperative phase.

INTRODUCTION

Dental restorative procedures often result in postoperative pain (Silvestri 1977; Piperno 1982; Grover 1984). Postoperative complaints resulting from thermal stimuli on teeth, i.e., hypersensitivity, has been investigated by Miller (1984). The sensory response time (SRT) was recorded both before, and one week after treatment. After recording the SRT of a tooth restored with an amalgam inlay, the restoration was replaced by an amalgam inlay with a cement base. The presence of the base reduced the hypersensitivity. However, in all 14 cases investigated the thermal sensitivity was greater than the pre-operative response.

Temporary postoperative hypersensitivity to thermal stimuli is not only the result of operative procedures, but may also be due to the high thermal diffusivity of metallic restorative materials, and may be related to the size of the restoration. Therefore, to gain insight into this phenomenon, thermal behavior of restored teeth has been studied by analyzing the temperature field in unrestored and in restored teeth with various cavity geometry and various restorative materials.

Thermal analysis can be carried out by calculating the temperature field in a theoretical model using Finite Element Analysis (FEA) (Lloyd 1978a; Wright 1978; Takahashi 1982; chapter 4.1). In a recent study (chapter 4.1), the temperature distribution in human teeth restored with an occlusal amalgam and various cement bases has been analyzed. The maximum interface temperature within the tooth model restored with amalgam only was 38.4^{o}C. The presence of a 0.5 mm thick cement base of calcium hydroxide ($Ca(OH)_2$) below this occlusal amalgam restoration was found to reduce the maximum temperature to 0.4^{o}C at the restoration-dentin interface. However, the calculations only related to identical occlusal restorations, and the affect of the presence of more complex restorations is in question.

The aim of this study is to analyze the influence of the cavity geometry on the temperature distribution within the underlying tissue, caused by a certain prescribed thermal load at the outer surface. Therefore, the heat transmission was analyzed to compare two extreme clinical situations: (a) an amalgam buildup representing the ultimate cavity geometry, and (b) a standardized occlusal restoration as a minimal design.

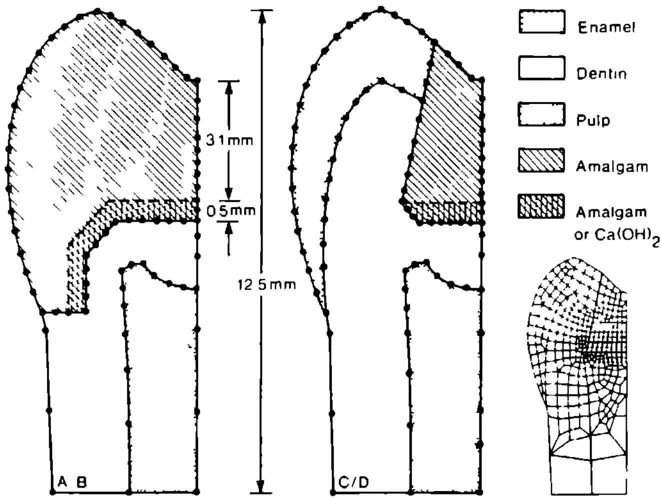

Fig 1 - Models (A) Amalgam buildup; (B) as A with 0.5 mm thick $Ca(OH)_2$ base; (C) occlusal amalgam; (D) as C with 0.5 mm thick $Ca(OH)_2$ base. The model at the right represent the element mesh used.

MATERIALS AND METHODS

The axisymmetric tooth model developed by de Vree (chapter 3) was used for the thermal analysis. Calculations were carried out using the FE-method. The geometry of the axisymmetric tooth model was defined according to a buccolingual cross-section of a mandibular molar tooth. An element mesh (313 elements and 338 nodes) was made using quadrilateral ring elements with a bilinear temperature field.

Thermal analyses were carried out for 4 models (Fig 1): (A) a molar with an amalgam buildup (i.e., the anatomical crown replaced with amalgam); (B) the same as A, but with a 0.5 mm thick cement base of $Ca(OH)_2$; (C) an identical tooth model with an occlusal cavity (depth 3.6 mm) restored with amalgam; and (D) the same as C, but with a 0.5 mm thick $Ca(OH)_2$ cement base.

The assumptions required for completion of the mathematical model as well as the thermal loading conditions were as follows (chapter 4.1): no radiation; no heat transport in tangential direction; continuous temperature distribution and heat flow at the internal interfaces; homogeneous, isotropic and temperature independent material properties; an initial temperature (T_o) of 35.2°C; prescribed heat flux through the coronal surface; and the temperature is prescribed at the subgingival cervical part of the root (T equals T_o).

The thermal load, applied to the coronal part of the outer surface, simulated the application of a warm liquid which is declining linearly in temperature from 60.0 to 35.2°C in 10 sec, and then remains constant during the rest of the time (20 sec). The heat transfer coefficient of the environment was given the value of 5.10^2 J/m^2s^oC (Jacobs 1973).

Under the given conditions, the four models have been analyzed comparatively for their effects on the temperature field within the models as a result of the different cavity geometries, and the presence of a cement base.

RESULTS

Figure 2 shows two plots of the temperature field near the pulp and the restoration of the models A/B and C/D 6.4 sec after initiation of the heat load. The direction of the heat flow - perpendicular to the isotherms - is hardly influenced by the change in cavity geometry. The density of isotherms and their corresponding temperature levels within one material provide information on the rate of temperature rise. Regarding the underlying dentin (i.e., the dentin between the top of the pulpo-dentinal juction (PDJ) and cavity floor) the presence of an amalgam buildup (models A and B) showed five isotherms corresponding to a temperature range of 39.2 to 41.6°C. The models with an occlusal restoration (models C and D) showed only three isotherms with a temperature range from 36.8 to 38.0°C. Thus, in the case of amalgam buildup, the rate of temperature rise increased and was approximately twice that observed with the occlusal restoration.

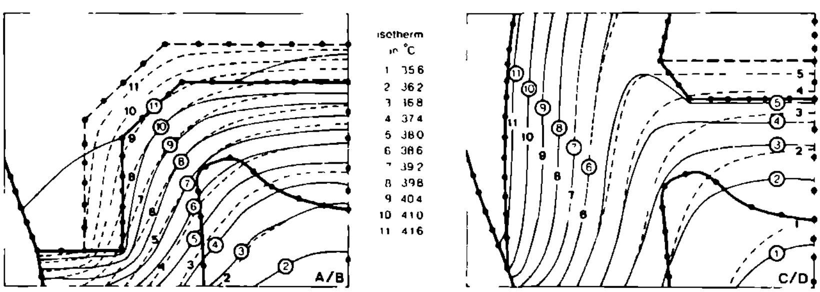

Fig 2 - The isotherms in the vicinity of the pulp and restoration of the models A/B and C/D on 6.4 sec after initiation of the heat flux. The solid lines with the encircled numbers 1 to 11 represent the isotherms within the models restored with amalgam only (A and C). The dotted lines with the bold numbers 1 to 11 represent the isotherms within the models restored with amalgam and a base (B and D).

In Figure 3, the correlation between temperature and time at node R is shown for the restoration-dentin interface during an elapsed time of 30 sec. Under the given conditions, the maximum temperature was 41.7°C for model A, 40.0°C for model B, 38.4°C for model C, and 38.0°C for model D; maximum temperatures were reached in 6.2, 6.2, 9.9, and 14.2 sec, respectively. This maximum rise in temperature for the models A and B were 3.3°C and 2.0°C, greater than those those for models C and D, respectively.

The maximum temperature increase at the border of the PDJ was calculated for node P and is shown in Figure 4 as a function of time. The maximum temperature was reached in the models A and B in 9.7 sec and in the models C and D in 14.2 sec. The maximum temperature were 39.2°C for model A, 38.4°C for model B, 37.4°C for model C, and 37.3°C for model D. The amalgam buildup without (A) and with (B) a cement base exhibited temperature increases at node P at 1.8°C and 1.1°C, greater than those observed in the models C and D, respectively.

DISCUSSION

To study changes in sensory response time (SRT) after dental treatment, in vivo experiments are necessary. The study by Miller (1984) provides information about thermal sensitivity and related SRT in different clinical situations. However, no information could be found about the temperature in the pulp of vital teeth or of the influence of the inter- and intrapatient variation on SRT. Although in vivo experiments concerning thermal behavior in restored teeth will be confined for practical reasons to temperature measurements within or at the border of restorations, a combination of experimental results and data from related identical theoretical research using FEA may contribute towards a better understanding of the unknown correlation between thermal processes and SRT.

Numerical mathematical methods provide calculated data about temperatures within tooth models of identical geometry, but various cavity design and restorative materials. However, due to lack of knowledge, some assumptions have to be made. For instance, only limited data could be derived from literature regarding thermal fluctuation in the environment of teeth while drinking liquids (Rothwell 1959; Plant 1974). Therefore, the thermal loading conditions and in particular those pertaining to the transient change in temperature during the consumption of liquids require further investigation.

Other assumptions to be verified in future studies relate to the conditions at the internal interfaces of the dental tissues and the restorative materials. The presence of cavity varnish and/or gaps at these interfaces can be accounted for by introducing a certain coefficient of heat transfer across these thin layers.

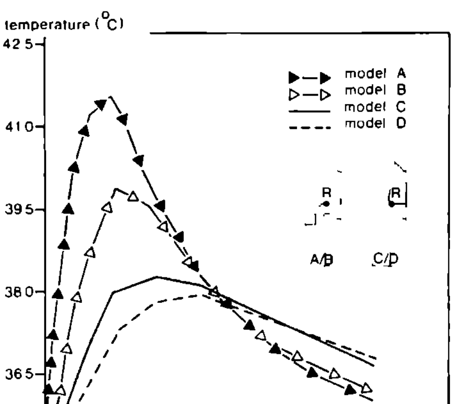

Fig 3 - The temperature change at node R in the models A, B, C, and D over a period of 30 sec after initiation of the heat flux.

The assumptions made limit the conclusions. However, in a recent study by Spierings (1986), the first results of a comparitive clinical experiment appear to confirm with those obtained from the model chosen. It is suggested that the results of this study may be considered to be representative of events in the clinical situation.

The temperature curve for model A at node R (Fig 3) was identical for all the nodes at this interface. By using a cement base (model B) the reduction in temperature increase at the restoration-dentin interface and at the PDJ was 1.7°C and 0.8°C, respectively. By reaching the maximum temperature in the same time period (6.2 sec), the cement base also influenced the rate in temperature rise as indicated by the slope of the curves in Figure 3. However, according to Miller (1984) and Silvestri (1977), the thermal sensitivity is still present despite the use of a $Ca(OH)_2$ base. These findings, combined with the thermal data reported here suggest that the application of a 0.5 mm cement base cannot be regarded as effective in protecting the dentin against thermal stimuli postoperatively.

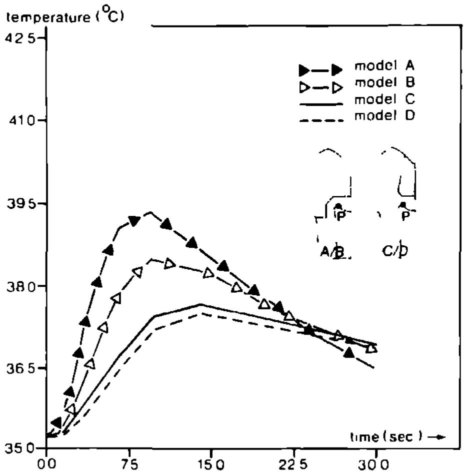

Fig 4 - The temperature change at node P in the models A, B, C, and D over a period of 30 sec after initiation of the heat flux.

According to Braden (1964b) and Harper (1980), a cement base of 0.75 mm would be more appropriate. In a pilot study the influence of 0.75 mm thick base of a reinforced zinc oxide-eugenol (ZOE) has been analyzed under the same loading conditions used in this study. Increasing the thickness of a ZOE base from 0.5 to 0.75 mm resulted in reductions of 0.2°C and 0.1°C in the temperature increase at the restoration-dentin interface and PDJ respectively. The clinical significance of these minor reductions in terms of postoperative discomfort are unknown.

Comparing our results with other studies, it must be realized that the temperatures in this study are directly related to the model chosen. The thermal loading conditions here are transient in contrast to the constant temperature conditions used in in vivo and in vitro measurements of thermal diffusion of various materials used in restorations (Voth 1966; Plant 1974; Tibbetts 1976; Harper 1980). Only Tibbetts (1976) described an in vivo measurement at the restoration-dentin interface of an extensive occlusal restoration during the consumption of a hot liquid. Their results appear to correspond with the data obtained for the models C and D, assuming a heat transfer coefficient of 5.10^4 J/m^2s^oC (chapter 4.1)

In a previous study (chapter 4.1), an unrestored tooth model was analyzed. The maximum temperature of 37.0°C at the PDJ was reached in 18.5 sec. Comparing this data to obtained in the present study, it is concluded that not only the temperature increase, but also the rate in temperature rise, could be an important factor in the reaction mechanism of the pulp to thermal stimuli. Trowbridge (1980) also related both temperature increase and rate in temperature rise to the dentinal fluid velocity and the sensory response.

The thermal diffusivity of amalgam and the geometry of the restored cavity affect the thermal load at the tooth tissues. However, the underlying tissue has a limited recuperative potential. If the pulp has already been irritated by caries, even a moderate amount of heat trauma may be fatal following local necrosis (Nyborg 1968). Notwithstanding the reported results, the use of a cement base has to be recommended in order to limit thermal traumata and to promote the production of reparative dentin.

Typical thermal loads result in atraumatic thermal behavior. Although a small restoration (like models C and D) might cause postoperative discomfort in 19% of cases when exposed to hot stimuli (Silvestri 1977), the changes in pulp tissues will be recuperated within 91 days (Zach 1965).

According to Zach (1965), a temperature increase of 5.6^{o}C in the pulp of a sound tooth leads to a necrotic pulp in 15% of all cases. By contrast, a temperature increase of 2.2^{o}C does not cause pulpal necrosis. Assuming a linear increase of occurence of necrosis within this temperature range, it follows that a temperature increase at the PDJ of 3.2^{o}C as in model B or 4^{o}C as in model A may induce necrosis in 4% and 8% of cases, respectively. However, in a restored tooth the pulp tissue has often been irritated by a caries lesion and subsequent dental procedures. In such circumstances the percentages of 4% and 8% probably represent minimum values.

Future research will be directed towards the study of the insulating capacity of new materials such as posterior composites and glass-ionomer cements.

5.1 THERMAL FLUCTUATIONS IN THE ORAL CAVITY DURING CONSUMPTION OF HOT AND COOL DRINKS*

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ABSTRACT - In view of research of heat transport in restored teeth, the ambient temperature of teeth has been recorded during the consumption of hot (65°C) and cold (5°C) liquids using a copper-constantan thermocouple.

The mean ambient initial temperatures were 36.1 ± 0.7°C for incisors and 36.4 ± 0.5°C for molars. The test liquids caused a mean maximum temperature change of 22.4 ± 3.1°C near incisors and 19.5 ± 3.1°C near molars.

The consumption of 1 draught liquid resulted in a characteristic temperature curve which has been described mathematically. The error function found comply with all recorded temperatures. The function can be helpful in simulation studies where the temperature distribution in teeth is studied by means of a finite element analysis.

INTRODUCTION

The temperature in the oral cavity is the overall result of complex physiological, physical and chemical processes. Thermal changes are mainly the result of physical processes related to breathing and food consumption. Extreme temperatures may lead to thermal irritation and subsequent damage of the hard dental tissues and pulp tissue. Specifically, the intake of food is indicated as being responsible for these phenomena (Langeland 1960; Zach 1965; Brown 1972; Lloyd 1978a). Moreover, the replacement of dental tissue by metal restorative material might result in higher/lower temperatures at the cavity walls and underlying tissues. Therefore, it is important to know the temperature-time relation at the outer surface of teeth under various physiological circumstances. The knowledge of this relation is necessary to study heat transport in restored teeth (chapters 3 and 4).

Rothwell (1959) recorded temperature fluctuations related to food intake in the oral cavity by placing a thermistor interproximally as well as at the occlusal side of the upper and lower second premolar and first molar. The recorded temperature during a standardized hot meal (65-70°C) and ice (3-8°C) varied between 45-48°C and 5-10°C, respectively. Gräf (1960) studied temperature changes within the enamel during the consumption of a test dinner. The various parts of the meal had a different temperature: 5°C (drinks), 75°C (soup and potato's) and -7°C (ice). Hot and cold consumptions were alternated. Recordings with a thermoprobe within the enamel at 0.5 mm from the outer surface resulted in a minimum temperature of 16°C and a maximum temperature of 45°C. On the occlusal side of an artificial lower left second premolar and first molar in a unisadle wax plate, Plant (1974) found maximum temperatures of 53.3°C, 50.0°C, and 47.0°C resulting from the consumption of coffee at a temperature of 63.5°C, 58.0°C, and 55.0°C, respectively.

The reported studies pay attention to minimum and maximum temperatures at the surface of teeth or in the enamel as a result of food intake with different temperatures. However, the reported values do not give any clue about the temperature as a function of time.

The aim of this in vivo study is to obtain the lacking information by recording temperatures in the environment of teeth as a function of time during and after the consumption of hot and cold liquids.

A thermocouple has been used to measure the temperature in the oral cavity. The thermosensitive part is tiny and a disturbance of the deglutition process is not probable. The instrument used is considered to be accurate and sensitive, which suits this study.

MATERIALS AND METHODS

A copper-constantan thermocouple was used with a wire diameter of 0.2 mm each. The electric potential in the thermocouple, consisting of reference side A and measuring side B, was recorded by a pen recorder (Kipp micrograph BD 121-E606, Delft, The Netherlands). The reference side was immersed in a waterbath with thermostat (Thermomix^{R} 1480, B. Braun Melsungen AG, Melsungen, FRG) ensuring a constant temperature level of 40°C. Figure 1 shows the position of the thermocouple during the experiment.

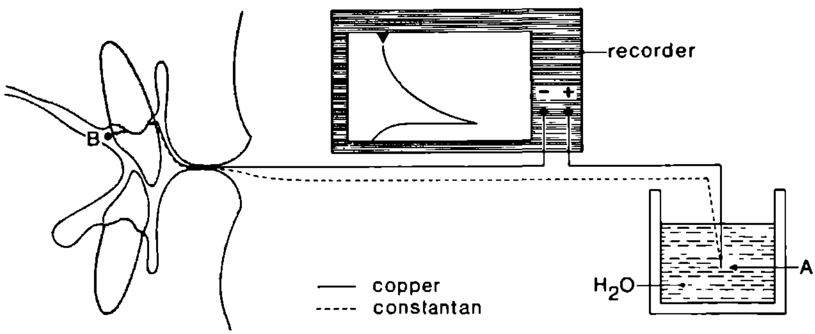

Fig 1 - Schematic representation of the test situation. A and B are the reference and measuring side of the thermocouple, respectively.

In a pilot study it was found that the most extreme temperatures were measured at the palatal side of the tooth. All experiments were therefore carried out with a thermoelement at the palatal side of an upper first incisor and first molar.

To study a possible additive effect of successive draughts, all measurements were carried out for one draught and for three consecutive draughts.

Four subjects (2 men, 2 women) in the age of 20-22 years participated in the experiment, two of them twice. They all received beforehand written information about the experiment. The experiments were always performed at the same time in the morning; the mean room temperature was 25.9 $\pm$ 0.2°C. The body temperature of each subject was registered sublingually at the start and finish of each series of experiments.

The procedure was as follows. The waterbath was heated to 40°C. During heating time (15 min) the subject received the last instructions, while acclimatizing. The mouth was only opened for drinking the test liquids. After calibrating the thermocouple, the sublingual temperature was recorded first. Then the thermocouple was fixed at the palatal side of the upper right central incisor. The sublingual and initial temperatures were registered after the recorded temperature was constant for at least 5 min. The temperature of the room, the waterbath, and the test liquid was checked before the experiment was started by means of a mercury-thermometer. The intra-oral temperature was recorded 5 min from the start (i.e., immediately after opening the mouth). Each series consisted of 4 experiments: 1 and 3 draughts of the hot liquid (65°C) and then 1 and 3 draughts of the cold liquid (5°C) were given successively. After each experiment the subject had to wait untill the local temperature returned to the starting level. After these 4 experiments, the thermoelement was fixed at the palatal side of the upper first molar. The same series was repeated. Finally the body temperature was recorded again.

RESULTS

Experimental

The mean initial sublingual temperature of the test persons was found to be 36.3 ± 0.9°C, being the same at the end of the experiments.

In Table 1, both the local initial temperature at the palatal side of the upper right first incisor and first molar are given as well as the maximum and minimum temperatures recorded during the consumption of hot and cold drinks. The mean ambient initial temperatures were 36.1 ± 0.7°C and 36.4 ± 0.5°C for incisors and molars, respectively. The hot test liquid (65.0°C) caused a mean maximum temperature of 56.5 ± 3.2°C near the incisor and 55.2 ± 4.8°C near the molar. Drinking the cold liquid (5.0°C) resulted in a recorded mean minimum temperature for the incisor and the molar of 12.0 ±

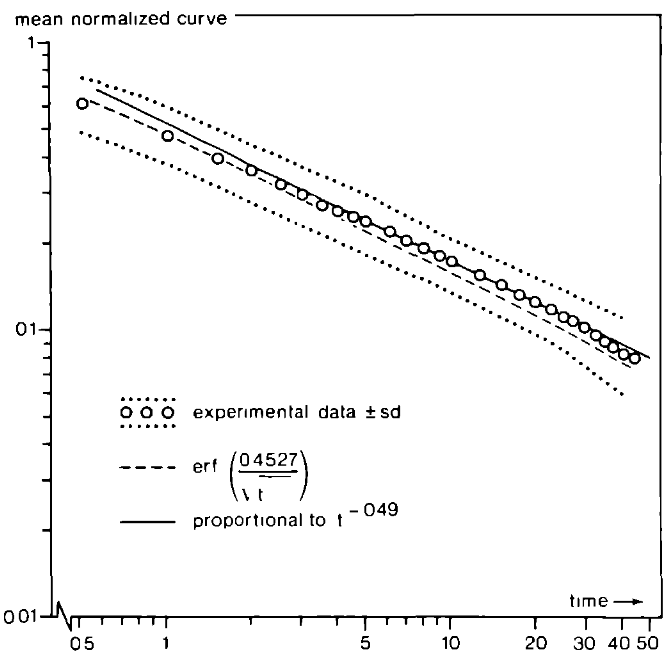

Fig 2 - Mean normalized temperature vs time t including the standard deviation, the function $t^{-0.49}$, and the described error function.

Table 1: Temperature recordings* at the palatal side of the upper first incisor (I) and the upper first molar (M) in ^{o}C.

testperson	A		B		C		D		Bb		Dd		mean temperature (sd)	
location	I	M	I	M	I	M	I	M	I	M	I	M	I	M
hot liquid (65^{o}C)														
$T_{initial}$	36.9	36.7	36.3	36.5	36.6	36.7	35.5	36.2	35.5	35.9	34.9	35.4	36.0(0.8)	36.2(0.5)
T_{max}	53.4	59.5	60.5	57.6	58.8	53.5	58.6	53.4	53.5	60.0	54.1	47.4	56.5(3.2)	55.2(4.8)
cold liquid (5^{o}C)														
$T_{initial}$	36.7	37.0	36.6	37.0	36.7	36.8	35.2	36.6	35.6	35.8	36.3	36.3	36.2(0.6)	36.6(0.5)
T_{min}	16.9	15.7	12.5	15.0	10.3	16.8	10.2	16.3	8.7	15.7	13.5	20.5	12.0(3.0)	16.7(2.0)

* Standard deviation 0.9^{o}C

$3.0^{o}C$ and $16.7 \pm 2.0^{o}C$, respectively. Apart from the sign, no significant differences were found between the temperature changes as a result of equal thermal load above or below the body temperature.

The second series of experiments (persons B and D, labeled Bb and Dd in Table 1) resulted in a recorded initial temperature lower than in the first run. The maximum temperature at the palatal side of the upper first molar is in experiment Dd much lower as compared to the mean maximum temperature.

In all experiments, the measured temperature did not return to the initial temperature within the recording time of 5 min. The experiment with 3 draughts of liquid in succession showed an additive effect only at the upper right first incisor during consumption of the cold liquid. The time lapse between the first and last high/low temperature varied from 3.2 to 10.5 sec. These results were not considered to be significantly different from the results obtained after one draught. Therefore, the results of these recordings are not represented.

With respect to the recordings of the consumption of one draught, a characteristic temperature-time curve was noticed. After normalizing the curves by using the formula:

$$Q = \frac{T - T_{body}}{T_{max} - T_{body}} \qquad \text{time scale such that } T_{(o)} = T_{max}$$

where Q is the normalized temperature of the recorded temperature T, the mean normalized curve has been calculated for all recordings separatedly as well as collectively for the experiments with cool/hot drink. Figure 2 shows on a log scale the mean normalized temperature versus time, resulting in a fairly straight line. No significant difference was found between the mean normalized curve as illustrated and the mean normalized curves calculated with the data of separated experiments (cool/hot). The slope of this line (- 0.49) is important for the theoretical model to be described below.

Theoretical model

After swallowing the liquid, conduction seems to be the dominating heat transfer process. Analyzing the experimental data, it was found that the temperature decline was more or less proportional to $t^{-0.5}$ (Fig 2); a time dependency typically encountered in problems where heat conduction takes place in only one spatial dimension. However, the oral cavity and surrounding tissues have a three-dimensional shape. Therefore, the experimental results suggest that the heat transfer problem at hand, is, to some extent, a spherically symmetric one.

A mathematical method was developed to obtain a suitable function for fitting the experimental data. It is based on the following assumptions:

a. The oral cavity is a vacuous sphere (radius r_1, center at the origin of the coordinate system) surrounded by an infinite medium with homogeneous and isotropic thermal properties (Fig 3);
b. Prior to drinking, the temperature of the medium is uniform and equal to the body temperature T_{body};
c. The passage of one gulp of liquid through the oral cavity raises the temperature initially, and significantly above the body temperature only in a relatively thin layer (thickness d) around the afore mentioned sphere. This initial temperature distribution is spherically symmetric (Figs 3 and 4);
d. After swallowing the liquid, the dominating heat transfer process is conduction. In this period, the net heat flux through the surface of the sphere due to radiation is zero;

e. The recorded temperature equals the temperature of the surface of the sphere.

By virtue of these assumptions the temperature T at any point depends only on time and on the distance r of that point to the origin of the coordinate system, that is $T = T(r,t)$.

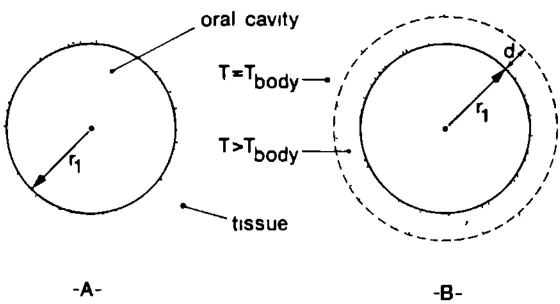

Fig 3 - Idealized shape of the oral cavity before (A) and immediately after drinking (B). T = temperature; r_1 = radius of the sphere; d = thickness of initially heated layer.

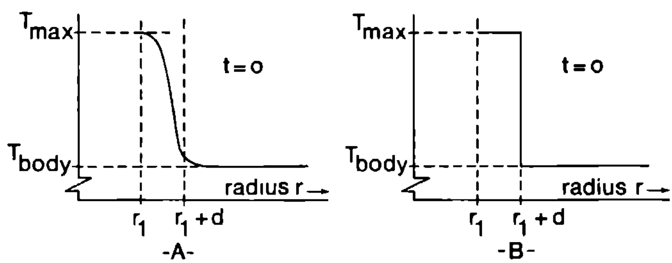

Fig 4 - (A) representation of the initial temperature distribution ($T = 0$); and (B) the approximated initial temperature distribution (steplike function).

Define the dimensionless quantities:

$\varphi = (T - T_{body})/T_r$; $s := r/r_1$; and $\tau := a.t/r_1^2$

in which T_r is some suitable reference temperature and 'a' the thermal diffusivity. The mathematical description of the heat conduction problem after swallowing one gulp liquid reads then as follows:

(1) $$\frac{\partial\varphi}{\partial\tau} - \left(\frac{\partial^2\varphi}{\partial s^2} + \frac{2}{s} \cdot \frac{\partial\varphi}{\partial s} \right) = 0 \qquad \tau \geq 0 \qquad s \geq 1$$

(2) $\tau = 0$ $\quad \varphi(s,0) = \psi(s)$ $\qquad$ (ψ: prescribed initial temperature)

(3) $s \downarrow 1$ $\quad \frac{\partial\varphi}{\partial s} = 0$ $\qquad$ (boundary condition: heat flux through surface = 0)

(4) $\varphi \to 0$ if $s \to \infty$ and/or $\tau \to \infty$

The requirement (4) is necessary to ensure that infinitely far from the oral cavity the temperature equals T_{body} and that over time the temperature at any point eventually should reach the value T_{body}.

In the appendix it is shown that the solution for the surface temperature $\varphi(1,\tau)$ is

$$(5)\quad \varphi(1,\tau) = \frac{1}{\sqrt{\pi\tau}} \int_0^\infty \exp\frac{-u^2}{4\tau}\left[(1+u).b(u) - e^{-u}\int_0^\infty (1+v).b(v).e^v.dv\right]du$$

The function b is determined by the prescribed initial temperature ψ. Setting $u = s - 1$ it holds that $b(u) := \psi(1 + u)$. Approximating the initial temperature distribution (Fig 4) by a steplike function (Fig 4) and setting $D = d/r_1$ and $b_o = (T_{max} - T_{body})/T_r$ one obtains for the approximated function b(u)

$$(6)\quad b(u) = \begin{cases} b_o & \text{for } 0 \leq u \leq D \\ 0 & \text{for } u > D \end{cases}$$

Substituting (6) into (5) yields

$$(7)\quad \varphi(i,\tau) = b_o\left[\underbrace{\mathrm{erf}\left(\frac{D}{2\sqrt{\tau}}\right)}_{\text{term } T_1} - \underbrace{D.e^{D+\tau}\left\{1 - \mathrm{erf}\left(\frac{D}{2\sqrt{\tau}} + \sqrt{\tau}\right)\right\}}_{\text{term } T_2}\right]$$

This is the solution to the equations (1)-(4) for $s \downarrow 1$ (surface), and b(u) defined by (6).

Analysis of the expression (7) reveals that the term T_1 in the asymptotic representation of $\varphi(1,\tau)$ is proportional to $\tau^{-1/2}$. Obviously, the model will yield qualitatively the same time dependence as that found for the experimental data.

From equation (7) a simple mathematical function can be derived suitable for describing the experimental data. The relative magnitude of terms T_1 and T_2 can be estimated with the aid of the inequality:

$$\left|\frac{T_2}{T_1}\right| \leq \frac{D\left(1 - \mathrm{erf}\left(\frac{D}{2\sqrt{\tau}}\right)\right)}{\mathrm{erf}\left(\frac{D}{2\sqrt{\tau}}\right)}$$

As the upper bound tends to zero if $\tau \rightarrow 0$ it follows that, for sufficiently small τ, term T_2 can be neglected. Therefore, the mean normalized experimental curve was described (see previous section) with the aid of the error function, i.e., $\mathrm{erf}(D/2\sqrt{\tau})$. It was found that $\mathrm{erf}(0.453/\sqrt{t})$ describes the experimental data sufficiently accurately (Fig 2) over a period of 40 sec after swallowing the liquid.

DISCUSSION

The time lapse for reaching the maximum or minimum temperatures is 1.0 ± 0.5 sec. So, the full scale response time of the pen recorder could interfere with the results, despite the small heat capacity of the thermocouple. However, the pen recorder used has a response time, i.e. full scale travel within 1 sec, which is shorter than the time lapse mentioned above.

The local initial temperatures at the palatal side of the upper right first incisor and first molar were found to be almost equal to the sublingual temperature. This rather high initial temperature measured with closed mouth in resting position, may possibly have been caused by the tongue touching the thermocouple. With respect to the extreme temperatures near the palatal side of the incisor and molar regions a certain difference between the measuring sites is observable, see Table 1. However, this difference does not seem to be systematic.

The variation in maximum and minimum temperature can be attributed to several factors. The size of the draught was not standardized to avoid an imposed deglutition, which might lead to an extremely long stay of the liquid in the oral cavity. The deglutition is a variable factor itself by which the direction of the liquid flow varies as well (Dijkman 1973). Furthermore, according to Boehm (1972) the ambient temperature, humidity and dryness of the oral cavity affects the temperature field of the cavity upon opening of the mouth. The difference between the first and second run of the test persons B and D may be explained by these factors.

In comparison to the maximum temperatures reported by Rothwell (1959) and von Gräf (1960), the present temperatures show larger thermal fluctuations. These differences can be explained by use of a thermistor and thermoprobe; the measuring sites; and the heat transfer coefficient difference of solid food which results in smaller temperature fluctuations at the tooth surface than with liquids of the same temperature (Jacobs 1973). The present mean maximum temperatures are in agreement with the results of Plant (1974). No difference was found between the in the literature reported and present mean minimum temperatures.

Factors which might influence the temperature versus time curve are for instance, the time needed for one draught and its volumetric flow rate; or the heat loss which can affect the recorded temperatures by conduction of heat along the thermocouple leads. According to Braden (1964a), these heat losses are negligible. So, all recorded graphs can be considered as representative for the change in temperature in the environment of teeth for the given situation.

During the consumption of a drink, the physical processes are extremely complex. The change in the ambient temperature of teeth is quick compared to the change in temperature within teeth. Thus, the latter will react more on the surrounding temperature and on the temperature change, thereafter. For these reasons, we did not try to model the drinking process it self, but considered only the time period <u>after</u> swallowing the liquid. The recordings of this study do not give information about the temperatures in the oral cavity during the consumption of hot/cold liquids. Additional research is recommended for that purpose.

Figure 2 shows, that for larger values of t, the function $t^{-0.49}$ is as accurate as the described error function. However if $t \rightarrow 0$, the function $t^{-0.49}$ becomes infinite, whereas the function $\mathrm{erf}(0.453/\sqrt{t})$ remains bounded. Therefore, the error function is more suitable for the entire time period (0 - 40 sec).

For the given conditions, the error function describes the thermal loading at the outer surface of the dentition. In simulation studies, where the temperature distribution has been calculated with the help of a theoretical model (de Vree 1983), the temperature of the thermal load was assumed to be a linear function of time. The present results show a non-linear change in temperature. In further research, these findings will be helpful.

APPENDIX

Consider the mixed problem for the parabolic partial differential equation

$$\frac{\partial\varphi}{\partial\tau} - \left(\frac{\partial^2\varphi}{\partial s^2} + \frac{2}{s} \cdot \frac{\partial\varphi}{\partial s} \right) = 0 \qquad \tau \geq 0 \qquad s \geq 1 \tag{A.1}$$

Initial condition: $\tau = 0 \quad \varphi(s,0) = \psi(s)$ (A.2)
Boundary condition: $s = 1 \quad \partial\varphi/\partial s = 0$ (A.3)
Asymptotic behavior: $\varphi \rightarrow 0$ if $s \rightarrow \infty$ and/or $\tau \rightarrow \infty$ (A.4)

Using the method of separation of variables it is straight forward to show that the function

$$\varphi(s,\tau) := \frac{1}{s} \int_0^\infty A(\omega) \cdot e^{-\omega^2\tau} \left\{ \sin \omega(s-1) + \omega \cos \omega(s-1) \right\} d\omega \tag{A.5}$$

solves for arbitrary $A(\omega)$ the differential equation (A.1). Moreover it meets the boundary and asymptotic conditions. If $\tau = 0$, it follows from (A.5) that $A(\omega)$ is the solution of the integral equation

$$s\psi(s) = s\varphi(s,0) = \int_0^\infty A(\omega) \left\{ \sin \omega(s-1) + \omega \cos \omega(s-1) \right\} d\omega \tag{A.6}$$

To obtain $A(\omega)$ we first define $A(\omega)$ to be the Fourier-sine transform of some, as yet unknown, function $W(u)$,

$$W(u) := \int_0^\infty A(\omega) \cdot \sin \omega u \cdot d\omega \tag{A.7}$$

Differentiation of (A.7) yields

$$\frac{dW}{du} = \int_0^\infty A(\omega) \cdot \omega \cos \omega u \cdot d\omega \tag{A.8}$$

Setting $u = s - 1$ and $b(u) := \psi(1 + u)$ in equation (A.6) we obtain with the aid of (A.7) and (A.8) that $W(u)$ solves the linear inhomogeneous differential equation

$$\frac{dW}{du} + W = (1 + u) \cdot b(u)$$

The general solution to this problem reads

$$W(u) = e^{-u} \left\{ W_0 + \int_0^u dv\,(1+v) \cdot e^v \cdot b(v) \right\} \tag{A.9}$$

W_0: arbitrary constant
Since $\omega \cdot A(\omega)$ and dW/du are Fourier-cosine transforms of each other, we have

$$\omega \cdot A(\omega) = \frac{2}{\pi} \int_0^\infty \frac{dW(u)}{du} \cdot \cos \omega u \cdot du \tag{A.10}$$

For the surface temperature (φ at s=1), it follows from (A.5) and (A.10) that

$$\varphi(1,\tau) = \frac{1}{\sqrt{\pi\tau}} \int_0^\infty \exp \frac{-u^2}{4\tau} \cdot \frac{dW}{du} \cdot du \tag{A.11}$$

To derive (A.11) the indentity

$$\sqrt{\frac{\pi}{4\tau}} \exp \frac{-u^2}{4\tau} = \int_0^\infty \cos \omega u \cdot e^{-\omega^2\tau} d\omega$$

has been used (Gradshteyn 1980). Requiring that $\varphi(1,\tau) \cong 0$ whenever

$b \equiv 0$, one finds that $W_o = 0$ (this requirement expresses that $\varphi(1,\tau)$ remains zero whenever $\varphi(1,0) = 0$, because afterwards, that is $\tau > 0$, heat is not supplied nor removed). As

$$\frac{dW}{du} = (1+u).b(u) - W(u) \tag{A.12}$$

we finally find from (A.11), (A.12) and (A.9) that

$$\varphi(1,\tau) = \frac{1}{\sqrt{\pi\tau}} \int_0^\infty \exp\left(\frac{-u^2}{4\tau}\right) \left[(1+u).b(u) - e^{-u} \int_0^\infty (1+v).b(v).e^v.dv\right] du$$

is the solution to the problem (A.1) - (A.4).

5.2 THE THERMAL DIFFUSIVITY OF TWO REPLICA RESINS*

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* Dent Materials 1986 submitted for publication

ABSTRACT - Two replica resins, i.e., a polyester and an epoxy resin are examined on the value of thermal diffusivity and on their suitability for pouring replicas.

The first mixture of each resin was molded into cubic specimens in which a thermocouple was embedded centrally. The change in temperature was recorded at the surface as well as at the center of the specimen. The thermal diffusivity could be calculated directly from the data obtained.

The polyester resin proved however to be unsuitable for pouring replicas and was not considered further. Thus, the thermal diffusivity has been determined only for a 2nd and 3rd mixture of the epoxy resin. The mean thermal diffusivity of the 2nd mixture was 1.34 ± 0.06 $10^{-7} m^2/s$, whereas the value of the 3rd mixture was 1.40 ± 0.05 $10^{-7} m^2/s$.

The epoxy resin was found to be suitable for pouring replicas. The method used appears to be suitable for direct determination of the thermal diffusivity.

INTRODUCTION

Theoretical analysis of transient thermal phenomena in restored teeth has shown that besides the geometry and composition of teeth, the thermal diffusivity is also an important factor (Braden 1964b; chapter 4.1). The extent to which these and other parameters will cause different temperature patterns can be studied with physical and mathematical simulations. Because of the complex structure and the numerous variations of natural teeth, it is preferable to preform experimental studies on replicas composed of one material having a thermal diffusivity within the range of dental tissue.

The purpose of this study is to determine the thermal diffusivity of a polyester (Type GTS + MEK Harder (5%), Romar-Voss, Roggel, The Netherlands) and an epoxy resin (Araldite D + Harder HY 956 (20%), Ciba Geigy, Basel, Switserland) since these data are not reported in literature. Only limited data on thermal properties of similar products are available (Table 1). For comparison, values for dentin are given as well.

The thermal diffusivity 'a' of a material depends on 3 parameters. It can be expressed as:

a = <u>thermal conductivity</u> / <u>density</u> x <u>specific heat</u>

By determining these parameters, the thermal diffusivity can be calculated. However, it is rather difficult to determine the specific heat of poor conductors like resins. Therefore, a direct method to determine the thermal diffusivity is applied in this study.

In the literature three direct methods are described using 1 or 2 thermocouples. Minesaki (1983) embedded 2 thermocouples within a cuboid specimen at various distances from the thermal loading site. However, the

Table 1: Material properties

Material (reference)	Thermal conductivity $J/m.s^{o}C$	Density kg/m^3	Specific heat $J/kg^{o}C$	Thermal diffusivity m^2/s
Polyester resin* (Dominghaus 1979)	$7.04\ .10^{-1}$	$2.00\ .10^{3}$	$1.20\ .10^{3}$	$2.93.10^{-7}$
Epoxy resin# (Ciba-Geigy)	$5.83\ .10^{-1}$	$1.55\ .10^{3}$	?	?
Dentin (Craig 1961) (Brown 1970)	$5.84\ .10^{-1}$	$1.95\ .10^{3}$	$1.60\ .10^{3}$	$1.87.10^{-7}$

*Type 802; #Araldite D (including Harder HY 956/Filler DT 079)

interferences of these couples with the actual temperature can be disputed. Tibbetts (1976) and Voth (1966) fixed 2 thermocouples within a layered specimen at the interface of various materials. The discontinuous heat transport at the interface could result in a heat pattern in which the interference of the second thermocouple is negligible. Thus, this method is considered suitable for determining the thermal diffusivity through materials composed of different layers. Braden (1964b) and Civjan (1972) placed 1 thermocouple within their specimens to record the change in temperature relative to the surrounding. No thermal fluctuation of the mercury bath could be observed, since the cold junction of the thermocouple was placed in the same mercury bath in which the specimen was immersed. However, in a pilot study using the same method, we noticed a thermal fluctuation of the mercury. This phenomenon is in the present study controlled as a variable.

Thus far, the shapes of the specimen used to determine the thermal diffusivity of a material were cylindric (Watts 1981) or cubic. Since a cylindric shape is more suitable for materials which are usually mixed in small amounts (Watts 1981), cubic specimens were preferred in this study.

First, the thermal diffusivity will be determined for both resins using cubic specimens. Thereby, both resins will be tested for their suitability for pouring replicas. The most suitable resin will be chosen. For subsequent studies, a second and third mix of this resin will be molded into samples of a cuboid shape and into replicas of various geometry for future thermal parametric studies.

MATERIALS AND METHODS

Theory

Within an isotropic, homogeneous, rectangular, solid block with the dimensions 2L, 2H, and 2W, and with initial temperature T_o and constant surface temperature of B_o, the temperature depends on time t and position $\underline{x}$ = (x,y,z) according to Carslaw (1973):

$$T(\underline{x},t) - B_o = (T_o - B_o) \cdot \frac{64}{\pi^3} \sum_{l=o}^{\infty} \sum_{m=o}^{\infty} \sum_{n=o}^{\infty} \frac{(-1)^{l+m+n}}{(2l+1)(2m+1)(2n+1)} \cdot \cos\frac{(2l+1)\pi x}{2L} \cos\frac{(2m+1)\pi y}{2W} \cos\frac{(2n+1)\pi z}{2H} e^{-m_{l,m,n}t} \qquad (1)$$

where $m_{l,m,n} = \frac{a\pi^2}{4}\left[\frac{(2l+1)^2}{L^2} + \frac{(2m+1)^2}{W^2} + \frac{(2n+1)^2}{H^2}\right]$

Under the given conditions and for sufficiently large values of time, the temperature at any point within the block can be derived from the first term of equation (1). Hence equation (1) can be simplified to:

$$T(\underline{x},t) - B_o = (T_o - B_o).A.e^{-m_o t} \quad (2)$$

in which $A = \frac{64}{\pi^3} . \cos\frac{x}{2L} . \cos\frac{y}{2W} . \cos\frac{z}{2H}$

$$\text{and where } m_o = \frac{a\pi^2}{4} . \left[\frac{1}{L^2} + \frac{1}{W^2} + \frac{1}{H^2}\right] \quad (3)$$

The thermal diffusivity a is expressed in m^2s^{-1}.

If the temperature difference between the surface and central part of the specimen is known as well as between the initial temperature of the specimen and its surrounding, the logarithm of $(T(\underline{x},t) - B_o)/(T_o - B_o)$ can be plotted as a function of time. In formula:

$$\ln(T(\underline{x},t) - B_o) - \ln(T_o - B_o) = \ln A - m_o t \quad (4)$$

Apart from the initial stages, a straight line should be obtained for $\ln(T(\underline{x},t) - B_o)/(T_o - B_o)$ as a function of time. The slope of this straight line is $-m_o$:

$$-m_o = \frac{d[\ln(T(\underline{x},t) - B_o) - \ln(T_o - B_o)]}{dt} \quad (5)$$

As mentioned before, the temperature of the bath was not constant. Hence, temperature B is also considered as a fuction of time. In the appendix, it is shown that equation (5) satisfies for sufficiently large values of time, provided temperature B is considered a function of time B(t). Therefore, the thermal diffusivity of the cuboid specimens can be obtained by substitution of formula (5) in equation (3).

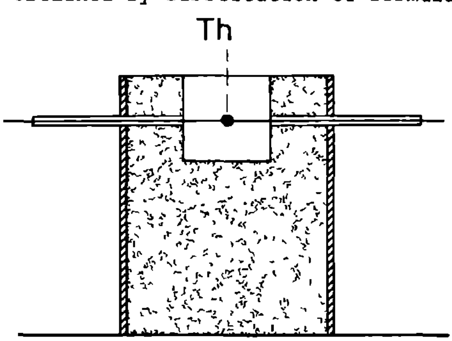

Fig 1 - The mold of a cuboid with the hot junction of the thermocouple positioned within its center.

Sample preparation

Cuboid specimens (0.0150 ± 0.0002 m) were prepared in silicone rubber molds held in sections of copper pipe (length 0.05 m, diameter 0.03 m). The hot junction of a copper-constantan thermocouple (A1-Cu-20/A1-Co-20, welded, Drijfhout BV, Amsterdam, The Netherlands), (42μV/^{o}C) has been embedded in its center (Fig 1).

Both resins - the polyester and the expoxy resin - were mixed according to the manufacturers' instructions, and molded into 3 samples each. To ensure complete curing, the resin specimens were dismantled 60 hours after initial mixing and kept at room temperature, thereafter.

After finishing the first set of measurements, a second and third mix of the preferred resin were molded, in order to assess the effect of possible variations in the mixture on the thermal diffusivity. The above mentioned preference was based on how well the resin could be poured.

Experimental measurements

The initial temperature of the specimen was 25.0^{o}C. When the temperature of the specimen was constant for at least 5 min, it was quickly transferred to a cold mercury bath of 0.0^{o}C. The specimen was kept in the bath for at least 3 min using an insulated metallic sample holder. This procedure was repeated 3 times for each sample of the first and 5 times for samples of the 2nd and 3rd set.

The thermocouple in the center of the specimen as well as a couple in the mercury bath were connected to a two-pen recorder (BD9, Kipp & Zonen, Delft, The Netherlands) via an ice reference junction. The thermocouple output (in mV) was recorded versus (vs) time. A chart speed of 500 mm/min allowed readings for every 6 sec. The readings were converted to ^{o}C, resulting in the temperature values ($T(\underline{x},t)$, $B(t)$, and T_o). By means of the method of the least squares, the slope as defined by equation (5) could be calculated.

The above procedure of data processing is time consuming and does not exclude reading errors. Therefore, during the third set of measurements, the thermocouple output was amplified and recorded vs time using a tape recorder (3968 Instrumentation Recorder, Hewlet-Packerd, Palo Alto, Ca 94304 USA). The data obtained were processed by analog to digital conversion that resulted in temperature plots of $\ln(T(\underline{x},t) - B(t))/(T_o - B(t))$ and directly calculated slopes ($-m_o$).

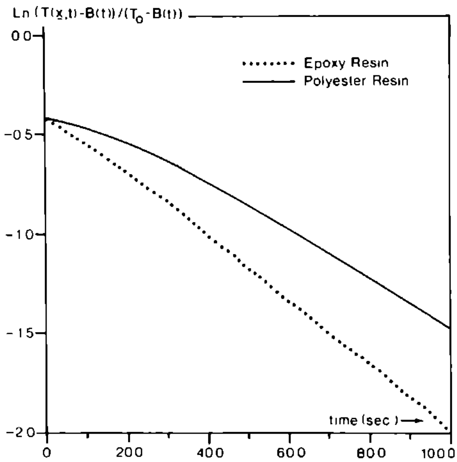

Fig 2 - Representative plot of $\ln(T(\underline{x},t) - B(t))/(T_o - B(t))$ for the epoxy and polyester resins versus time.

RESULTS

For each resin, a representative plot of $\ln(T(\underline{x},t) - B(t))/(T_o - B(t))$ vs time is shown in Figure 2. Similar curves were obtained for each individual specimen. Except for an initial transient period of about 30 sec, a straight line was obtained in accordance with the theory.

The mean diffusivity from 3 measurements of each specimen of the first set is 1.01 $\pm$ 0.05 $10^{-7} m^2/s$ for the polvester and 1.30 $\pm$ 0.10 $10^{-7} m^2/s$ for

Table 2: The mean thermal diffusivity for polyester and epoxy resin (n recordings per sample A, B and C).

		Thermal diffusivity 'a' (10^{-7} m^2/s)	
Sample	n	Polyester resin 1st mixture	Epoxy Resin 1st mixture
A	3	1.01 ± 0.03	1.31 ± 0.09
B	3	1.06 ± 0.07	1.28 ± 0.07
C	3	0.95 ± 0.04	1.32 ± 0.13
a ± sd		1.01 ± 0.05	1.30 ± 0.10

the epoxy resin (Table 2).

Although, both resins seem to be suitable for the construction of replicas, the epoxy resin was chosen. The results of the 2nd and 3rd mixture of epoxy resin are noted in Table 3. The data concern the mean thermal diffusivity of 5 measurements of each specimen.

DISCUSSION

The recorded thermal fluctuations in the mercury bath itself of 0.13 to 0.39°C allowed us to improve the experimental results by taking this thermal fluctuation into account. Instead of using $\ln(T(\underline{x},t) - B_o)/(T_o - B_o)$ vs time as per equation (5), the plot of $\ln(T(\underline{x},t) - B(t))/(T_o - B(t))$ as a function of time has been used. This has resulted in data which are 5% higher compared to the data following from the first function, i.e., assuming a constant mercury bath temperature. On the other hand, according to the Appendix, our results could be made even more accurately, provided $\ln \varphi(t)$ and $\ln A(t)$ were also taken into account.

The sample holder is more suitable for immersing the specimen into the mercury than the ceramic rod used by Civjan (1972). The sample holder used in this study causes a reduced mercury-resin interface, which is assumed to have a negligible influence on the heat flow.

The plots of Figure 2 are in agreement with the plots presented by Braden (1964b). Both plots show curved lines in the initial stages of about 30 sec. According to the theory, straight lines were obtained for large values of time. Therefore, the first part of the curve is excluded from the determination of the slopes of the plots.

Table 3: The mean thermal diffusivity for epoxy resin of n recordings per sample (A, B, and C)

		Thermal diffusivity 'a' (10^{-7} m^2/s) for Epoxy resin	
Sample	n	2nd mixture	3rd mixture
A	5	1.37 ± 0.08	1.27 ± 0.06
B	5	1.31 ± 0.03	1.52 ± 0.04
C	5	1.31 ± 0.06	1.41 ± 0.04
a ± sd		1.34 ± 0.06	1.40 ± 0.05

The difference between the data of the sets of epoxy resin (Tables 2 and 3) may be partly due to the inhomogeneity of the mixture, the amount of harder added and the incorporation of macroscopically invisible air-bubbles. Another reason could be the more accurate data processing as used for the third set.

Comparing the data in Table 1 with the present results, it can be seen that the polyester resin shows a relatively low value of thermal diffusivity compared to the one of dentin. According to the manufacturers' instructions polyester resin should be suitable for making replicas. However, in actual use the mixture had to be put under vacuum to avoid air-bubbles in the specimens. Moreover, the surface of the polyester resin did not completely set in the presence of oxygen, so a nitrogeneous surrounding had to be used. Therefore, the epoxy resin was chosen for pouring replicas.

The optimized method described here has proved to be suitable for direct determination of the thermal diffusivity of the given material.

APPENDIX

An isotropic, homogeneous, rectangular, solid block specimen with dimensions 2L, 2W, and 2H, and with temperature T_o at time t = 0, is placed in a bath with temperature B(t). It is reasonable to assume that the surface temperature of the specimen equals temperature B(t). The temperature at point $\underline{x}$ = (x,y,z) within the specimen is given by

$$T(\underline{x},t) - B(o) = (T_o - B(o)).\varphi(\underline{x},t) + \int_o^t \frac{dB(\tau)}{d} (1-\varphi(\underline{x},t-\tau)d\tau \quad (A.1)$$

in which $$\varphi(\underline{x},t) = \frac{64}{\pi^3} \sum_{l=o}^{\infty} \sum_{m=o}^{\infty} \sum_{n=o}^{\infty} \frac{(-1)^{l+m+n}}{(2l+1)(2m+1)(2n+1)} .\cos\frac{(2l+1)\pi x}{2L} \cos\frac{(2m+1)\pi y}{2W} \cos\frac{(2n+1)\pi z}{2H} e^{-m_{l,m,n}t}$$

and where $$m_{l,m,n} = \frac{a\pi^2}{4}\left[\frac{(2l+1)^2}{L^2} + \frac{(2m+1)^2}{W^2} + \frac{(2n+1)^2}{H^2}\right]$$

Substituting B(t) = B(o) + v(t) into equation (A.1) yields

$$T(\underline{x},t) = B(t) + (T_o - B(t)).\varphi(\underline{x},t) + v(t).\varphi(\underline{x},t) - \int_o^t \frac{dv(t)}{d\tau} \varphi(\underline{x},t-\tau)d\tau \quad (A.2)$$

$$\text{Hence,} \quad \frac{T(\underline{x},t) - B(t)}{T_o - B(t)} = \varphi(\underline{x},t).D(\underline{x},t) \quad (A.3)$$

where $D(\underline{x},t) = 1 + \frac{v(t)}{T_o - B(t)} - \frac{1}{T_o - B(t)} . \frac{1}{\varphi(\underline{x},t)} \int_o^\tau \frac{dv(\tau)}{d\tau} . \varphi(\underline{x},t-\tau)d\tau$

The logarithm of (A.3) can be written as

$$\ln(T(\underline{x},t) - B(t)) - \ln(T_o - B(t)) = \ln\varphi(\underline{x},t) + \ln D(\underline{x},t) \quad (A.4)$$

For the first 100 sec, it appeared that v(t) could be estimated by v(t) = b.t with $b = 2.10^{-3}$, whereas v(t) decreased again for t > 100 sec. Substituting v(t) = b.t in $D(\underline{x},t)$ yields

$$D(\underline{x},t) = 1 + \frac{b.t}{T_o - B(o) - b.t} - \frac{1}{T_o - B(o) - b.t} . \frac{1}{(\underline{x},t)} b(\Phi(o) - \Phi(t))$$

where $\Phi(t)$ is the primitive of $\varphi(\underline{x},t)$

For large enough values of time, e.g., t > 50 sec, $\varphi(\underline{x},t)$ can be approximated by the first term of $\varphi(\underline{x},t)$ (see A.2 and A.3). Neglecting the second and following terms of $\varphi(\underline{x},t)$ results only in a relative error of about 5% in $\ln\varphi(\underline{x},t)$. Now D(x,t) can also be estimated. It is found that $|\ln D(\underline{x},t)| < 0{,}01\ |\ln\varphi(\underline{x},t)|$ for t > 50 sec. This justifies the derivation of the thermal diffusivity 'a' from the expression for $\ln(T(\underline{x},t) - B(t))/(T_o - B(t))$ even when the temperature of the mercury bath is not exactly constant during the experiment.

5.3 DETERMINATION OF THE CONVECTIVE HEAT TRANSFER COEFFICIENT*

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* Dent Materials 1986 submitted for publication

ABSTRACT - The value of the convective heat transfer coefficient (htc) is determined under different loading conditions by using a computer aided method. The thermal load has been applied mathematically as well as physically to the coronal surface of an axisymmetric tooth model.

To justify the assumptions made for the mathematical tooth model, the medium mercury is used first. For all the other thermal loading conditions the medium water has been used. During the experiments the temperatures were recorded at one point within the resin model.

The temperatures within the theoretical model have been calculated according to the FEA. Comparison of the calculated data with the experimental recordings led firstly to a verification of the theoretical model, and secondly to a htc-value of 5.10^4 J/m^{2}s^{o}C.

From the findings of this study it can be concluded that this value of the htc satisfies for the experimental conditions used. However, additional in vivo experiments will be needed to verify the htc-value during the consumption of liquids.

INTRODUCTION

The convective heat transfer coefficient turned out to be an essential parameter in a study on temperature distribution in restored human teeth using a mathematical model (chapter 4.1). In convective heat transfer processes, the heat transfer coefficient (htc) represents the quantity of thermal energy transferred in a unit time at a fluid-solid interface of a unit area having a unit temperature difference. Its value is a complex function of the flow rate of the fluid, the fluids physical parameters, and the interface geometry of the solid into which or from which the thermal energy is transferred.

Enlarging the value of the htc results in an increase in temperature change of the underlying structures due to the faster energy transfer. In general, the heat transfer of a beverage will be larger in comparison with solid food of the same temperature (Jacobs 1973). So, the temperature changes in teeth are larger from drinking a hot beverage than eating hot solid food of the same temperature. Since we are interested in the worst case the thermal load in this study will be limited to fluid applications.

Jacobs (1973) investigated the heat transfer coefficient during application of a constant flow rate of the fluid. In the oral cavity, however, the flow rate is not constant while drinking a liquid. Moreover, the outer surface of the dentition is not necessarily uniformly wetted. Thus the value of the htc during the heat transfer may vary over the outer surface of the coronal part of the dentition. Determination of the htc-value for such a complex process might yield a mean value, which depends on the physical factors involved. In recent studies on temperature distribution in restored teeth (chapter 4), the value of htc given by Jacobs (1973) has been used. However, comparison of our results with the in vivo measurements of Tibbetts (1976) indicates that a larger value of htc might be more realistic (chapter 4.2). Since the value of the htc is an important parameter in studies concerning heat transport, and since only limited data on the value of htc could be derived from literature, it is the purpose of this study to determine its value under different loading conditions.

The complexity of and variation in human tooth structure make natural teeth unsuitable for standardized experiments. This points to the needs to use a simplified axisymmetric tooth model. Since this heat transfer process is complex in nature, it is rather difficult to determine the value of the htc in the laboratory alone. Consequently, a computer aided method was used. The different thermal loading situations were exerted in theoretical simulation models as well as in laboratory experiments. Combining the calculated data of the theoretical model with the results of the corresponding experiments will yield the value of htc for the given conditions.

MATERIAL AND METHODS

A mathematical as well as a physical model have been used to determine the value of the htc for water running along the coronal surface of an axisymmetric tooth. The geometry of each theoretical and experimental, axisymmetric tooth model was based on the bucco-palatal contours of a maxillary first premolar tooth, as derived from an X-ray. The physical model was made of an epoxy resin (Araldite D, Ciba, Basel, Switserland).

When one is drinking a liquid, only the coronal part of the tooth is involved. This situation comprises several physical parameters. It also introduces the question of whether the apical side must be kept at a constant temperature or has to be loaded similarly to the coronal part. Thus, this question has to be answered before experiments are carried out with water as the surrounding liquid, and before a comparison can be made to the corresponding theoretical model. Therefore, one experiment is performed with mercury as the medium, which has an infinite large value of htc. This implies that there is essentially no temperature gradient at the outer surface of the model (Carslaw 1973).

To determine the value of the htc, the following loading conditions were simulated: (A) A thermal load having a constant temperature of 0.0°C is applied at the coronal surface of the model for at least 60 sec, using mercury. The apical side of the model is not loaded; a constant temperature is prescribed at this surface, i.e., equal to the initial temperature of the model; (B) the same as A, however, stagnant water is used instead of mercury; (C) the same as B, however, the load is applied for 10 sec, and then the model is returned into an ambient condition of 30°C for the remainder of time (50 sec); and (D) a draught of 30 ml water at 0.0°C is simulated to determine a possible influence of running instead of stagnant water on the value of the htc.

For practical reasons, the initial temperature of the physical tooth model was 30.0°C. In the mathematical and experimental models, the temperatures caused by the various thermal loads have been calculated or recorded at one point. This point corresponds with the top of the pulpo-dentinal junction (Fig 1). The comparison of the theoretical data with the experimental results could lead to a verification of the theoretical model and to the determination of the htc-value.

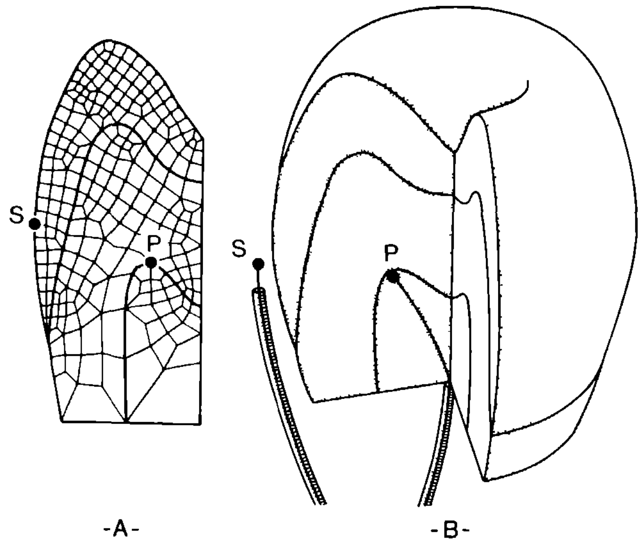

Fig 1 - (A) cross-section of the axisymmetric model, including the element mesh; and the nodes S and P corresponding the two thermocouples S and P of the physical model (B)

Mathematical approach

The calculations were carried out according to the finite element method. Therefore, an element mesh was made for the axisymmetric tooth model using quadrilateral ring elements with a bi-quadrilateral temperature field. The total number of elements and nodes was 238 and 769, respectively (Fig 1).

The material properties of the epoxy resin used were assumed to be isotropic, homogeneous, and temperature independent. Its assigned value of thermal diffusivity was 1.40 10^{-7} m^2/s (chapter 5.2).

To complete the mathematical model, the following assumptions were made: no radiation; no heat transport in tangential direction; and a prescribed heat flux through the surface. For loading situation A, a constant temperature of 30°C is prescribed at the apical side.

Using mercury as the medium, the htc was given the value of 5.10^{12} J/m^2s^oC, resulted in a tooth surface temperature equal to the temperature of the mercury. Calculations were made and the results were verified with those from the corresponding physical experiments.

After the verification of the modeling chosen, the value of htc will be fitted to match the calculated data with the results of the experiments, following from the given loading conditions B and C using water as the medium.

The value of the htc pertaining to load B, using water as the medium, was used to simulate load D. One draught of liquid was simulated as a function of time by the error function $erf(0.453/\sqrt{t})$ (chapter 5.1). The results of the modeling chosen were compared with the experimental data.

Experimental approach

To prepare a physical model, an epoxy resin was poured in a silicone rubber mold in which the hot junction of a copper-constantan thermocouple was located (Fig 1) at the point corresponding to P in the theoretical model. The resin was mixed according to the manufacturers' instructions.

For initial heating the model was stored for at least one hour in a hot-air stove at a constant temperature of 30.0°C.

A mercury bath was used to verify the theoretical model. The mercury was maintained at a loading temperature of 0.0°C by a jacket of ice water. The axisymmetric tooth model was immersed into the mercury bath at loading condition A as mentioned above.

The temperature of the water bath was maintained constant at 0.0°C, using ice-water. The experiments without running water (loads B and C) were carried out like the one with mercury; 30 ml of ice-water was used to simulate loading condition D (a draught of liquid). The water was applied on top of the tooth model.

During the various thermal loadings, the change of the ambient temperature and the temperature change within the model were recorded using a copper-constantan thermocouple. The reference junctions of the thermocouples were kept in ice-water. The thermocouple output was amplified and recorded versus time using a tape recorder. All experiments were repeated 5 times.

For comparison of the theoretical and experimental data of the loading conditions A and B, all temperature-time curves were plotted for $\ln(T(t)/T_o)$ as a function of time, where T is the measured temperature at time t, and T_o is the initial temperature of the model. Under the given loading conditions, the theoretical data had to correspond with the given results of the laboratory experiments.

During the transport of the physical model from air-stove to thermal loading place, its initial temperature will be lower than in the theoretical model. Therefore, all data obtained from the loading conditions C and D were normalized first, by dividing the calculated or recorded temperatures by the initial temperature of the model.

Table 1: Direction coefficient of $\ln(T(t)/T_o)$ and the corresponding heat transfer coefficient (htc) as result of the thermal loads A/B.

	thermal load (°C)		direction coefficient $\ln(T(t)/T_o)$		Value of htc
medium	theory	experiment	theory	experiment	J/m^2s^oC
A: mercury	0.0	0.7 ± 0.2	-0.023	-0.021 ± 0.001	∞
B: water	1.5	1.6 ± 0.4	-0.021	-0.019 ± 0.001	5.10^4

T(t) and T_o are temperatures at time t and t = 0, respectively.

RESULTS

The mean initial temperature of the physical tooth model was 29.3 ± 0.5°C. For each loading condition (A and B), the chosen thermal load value (theory) as well as the recorded average loading temperature (experimental) is given in Table 1. The modeling chosen in loading condition A fits the experimental results using mercury. Thus for loading condition B, using water as medium instead of mercury, the given value of htc appeared to be the best value of the parameter fitting the experimental data. Increasing this value had no significant influence on the calculated data at point P.

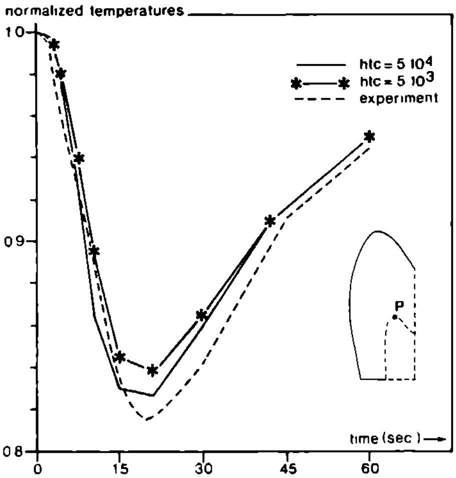

Fig 2 - The normalized temperatures for load C at point P in the experimental and theoretical model, having a heat transfer coefficient (htc) of 5.10^4 and 5.10^3 J/m^2s^oC during a period of 60 sec.

In Figure 2, the normalized temperatures at point P for load C are plotted as a function of time. The calculated temperatures at node P using a htc of 5.10^4 J/m^2s^oC are in agreement with the data obtained experimentally, whereas a smaller value of htc appeared to have a negative influence on the magnitude of the calculated data. Enlarging the htc-value had still no significant influence on the temperatures at node P. The thermal load caused at P a minimum temperature of 24.8°C in the theoretical model (i.e., htc is 5.10^4) and 24.1 ± 0.4°C within the physical replica.

The calculated and recorded normalized temperatures within the model (P) and at the surface of the model (S) as a result of load D, are depicted in Figure 3. Although point S at the surface of the theoretical model does not correspond exactly with the localization of the nearby thermocouple, the recorded and calculated data show close agreement. For point P, however, the initial stages of the curves show a difference in temperature change.

DISCUSSION

In this study, only one thermocouple was situated in the physical model. If more than one measuring site would have been present, more information could have been obtained about the temperatures at these sites and about the temperature gradient between these points. However, the use of more than one thermocouple in such a small model was not possible without influencing the temperature field. The use of only one thermocouple is therefore limiting the conclusions.

With respect to the temperatures at point P, all calculated results using FEA correspond with the experimental data using the given value of htc. However, enlarging the htc did not result in other calculated temperatures at P, whereas the temperature gradient to the surface increased. On the other hand, reduction of this value caused smaller temperature changes within the entire model (Fig 2). Thus, the value of htc obtained represents the lower bound for the given conditions. If the thermocouple was located near the surface, it could be disputed whether the results could inform us better about this parameter, specifically if the load is not uniformly applied to the coronal part of the model. Further research will be necessary to determine the 'mean' value of htc during the heat transfer process of drinking a beverage, leading to a more accurate figure for the temperature level at the surface of a tooth.

The temperature differences between the curves for node P as shown in Figure 3 can be attributed to the influence of running water on the value of htc, indicating that this value should be higher. In this study, however, the application of the liquid on top of the tooth model caused a uniform-wetted surface, whereas in vivo the wetting degree will not be uniform at the coronal part of teeth when drinking a beverage. Moreover, enlarging the value of htc did not have any influence on the temperatures at node P. Therefore, it is recommended that loading situation D be repeated in an in vivo experiment. These results could then be compared to the data obtained from the theoretical model simulating this loading condition. From those findings, it might be concluded whether the value of htc, i.e. 5.10^4, represents the 'mean' value.

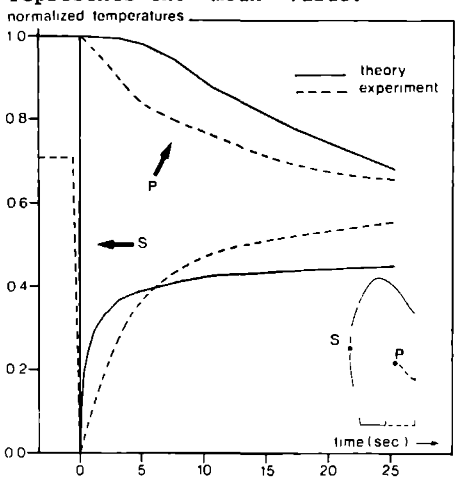

Fig 3 - The normalized temperatures for load D at the points S and P in respectively the theoretical and experimental model during a period of 25 sec after initiation of the heat flux. However, the thermocouple (S) was located near point S not in it.

The values of htc for liquids determined by Jacobs (1973) are lower than the value presented in this study. In his study, the fluid was poured over a small cylinder used as measuring object. It is in question whether in that situation the volumetric flow rate of 6.0 cm^3/sec resulted in a similar situation along the surface of the dentition while drinking a beverage. A smooth surface copper cylinder was used, whereas the tooth replica used in our study had a rougher surface. This may cause a different turbulent flow and therefore higher heat transfer for the rougher surfaced model resulting in an effectively larger wetted area for the courser model, which further enhances heat transfer even under no-flow conditions. Moreover, the flow velocity of liquids passing the oral cavity is not known from literature. These factors could be responsible for the reported differences. The large value of htc appears to correspond with the calculations on temperature distribution within restored teeth as reported before (chapter 4.2).

From the findings of our study, it can be concluded that (A) the htc-value of 5.10^4 J/m^2s^oC satisfies the experimental conditions used; and (B) additional in vivo experiments will be needed to determine the 'mean' value of the convective heat transfer coefficient during and after the consumption of liquids.

5.4 HEAT TRANSMISSION IN TEETH; VERIFICATION OF THEORETICAL MODELING BY IN VIVO EXPERIMENTS*

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ABSTRACT - A theoretical model simulating the conditions while drinking a liquid of a certain temperature, was compared with a similar in vivo experiment. Therefore, a replica of an axisymmetric tooth model has been used as experimental model for in vivo tests.

The temperature changes as result of one draught of a hot/cold liquid are recorded within the model as well as in the environment of the tooth model. These experimental data obtained were compared with the calculated results of the theoretical model as determined by the FEA.

The temperatures experimentally recorded appear to agree with the calculated results. It can be concluded that the assumptions, which have been made concerning the described thermal loading conditions, lead to a good approximation of the physical reality.

INTRODUCTION

In recent studies (chapter 4), transient heat transport problems within axisymmetric models of unrestored and various restored teeth were theoretically analyzed using Finite Element Analysis (FEA). Due to lack of detailed knowledge concerning the thermal load on teeth caused by a draught of liquid, the ambient temperature imposed by the liquid change was assumed linearly over time (chapter 3). To affirm this assumption, an in vivo experiment has been carried out to determine the temperature change in the oral cavity when drinking hot or cool beverages. However, the temperature-time curve found as result of this experiment appeared to agree with a non-linear function of time instead, i.e., an error function (chapter 5.1).

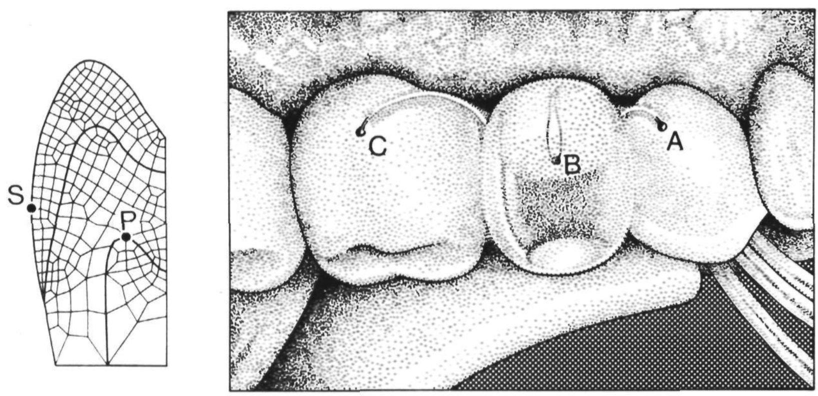
Besides the thermal load itself, the heat transfer coefficient (htc) is an important parameter in the thermal process of energy transport into teeth. Comparison of our results which concern temperatures at the restoration-dentin interface caused by a thermal load as determined by the FEA (chapter 4.1), with a similar in vivo experiment (Tibbetts 1976), does indicate that the value of this htc as reported by Jacobs (1973) did not correspond with this loading condition. This conclusion was further supported by the results obtained in a recent study in which the value of this coefficient was determined (chapter 5.3). However, it has to be realized that the given value of 5.10^4 J/m^2s^oC represents the value following from the simulation of the consumption of a draught of liquid in an <u>in vitro</u> experiment. To ensure that the presented value is also realistic under <u>in vivo</u> conditions, this value will be used and tested in this study.

Thus far, no justification was made for the theoretical model to calculate the temperature distribution with the FEA within teeth as a result of a thermal load which simulates the draught of hot or cool beverages. The purpose of this study is to verify the theoretical model by comparing it with an "in vivo" experiment under similar thermal conditions.

The use of vital teeth can not be justified for ethical reasons. Moreover, the diversity in geometry and composition of human teeth will result in different temperature distribution patterns within teeth. Therefore, a replica of an axisymmetric model is used as experimental model for these in vivo tests.

MATERIALS AND METHODS

The geometry of each theoretical and experimental, axisymmetric tooth model was based on the bucco-palatal contours of the three structures (enamel, dentin, and pulp) of a maxillary first premolar tooth, as derived from an X-ray. Because of the insulating effect of the environmental tissues on the root structure, and for practical reasons, the tooth model was restricted to 2 mm below the cemento-enamel junction (Fig 1).



—a— —b—

Fig 1 - (a) cross-section of the axisymmetric model, including the element mesh and the nodes S and P corresponding the three thermocouples A/C and B of the bridge in situ (b)

Theoretical approach

An element mesh (238 elements, 769 nodes) was made using quadrilateral ring elements with a bi-quadratic temperature field (Fig 1).

Thermal analysis was carried out for a tooth model composed of one material, i.e., epoxy resin (epoxy; Araldite D + Harder HY 956 20%, Ciba-Geigy, Basel, Switserland). The material properties were supposed to be isotropic, homogeneous, and temperature independent. The assigned value of the thermal diffusivity was 1.40 $10^{-7} m^2/s$ (chapter 5.2).

The following assumptions were made for the completion of the mathematical model (chapter 4): no radiation; no heat transport in tangential direction; and a prescribed heat flux through the coronal and cervical part of the model.

The thermal load applied to the axisymmetric tooth model, having an initial temperature of 35.2^{o}C, simulated the draught of a warm (60^{o}C) or cold (0^{o}C) liquid. In a recent study by Spierings (chapter 5.1), the temperature change in the environment of teeth caused by a draught of hot or cold liquid was found to agree with the error funtion, i.e., $erf(0.453/\sqrt{t})$. In this study, the thermal load will be simulated by this time dependent error function. The maximum/minimum ambient temperatures correspond with the temperatures recorded simultaneously during the in vivo tests, which are carried out to verify the theoretical modeling.

The heat transfer coefficient for energy transport from the liquid into the tooth model was assigned to the value 5.10^4 J/m^2s^oC (chapter 5.3). Since, the verification model was fixed as pontic, and because the value of htc represents a mean value, this coefficient was supposed to be applicable to the cervical part as well.

The calculations were carried out using FEA.

Experimental approach

The physical tooth model was an epoxy replica of the theoretical design. The replica was fixed as pontic in a bridge (Fig 1). During the test period, the bridge was situated in the upper premolar region of 2 test persons, who volunteered after informed consent about the tests to be carried out.

The thermal load consisted of a 30 ml draught hot (60^{o}C) or cold (0^{o}C) liquid. The liquid intakes were repeated 5 times each.

Three copper-constantan thermocouples were used to register the changes in temperature as a function of time within and outside the pontic during and after the drinking a hot or cool beverage. Since the presence of a thermoelement at the surface of the pontic would disturb the local heat transport into the pontic, two thermocouples were fixed at the palatal side of the adjacent crowns: A) mesial and C) distal of the pontic. The thermocouple (B) was embedded within the replica. The measuring sites A/C and B corresponded with the respective nodes S and P of the theoretical tooth model (Fig 1). The thermocouples were connected to a tape recorder via an ice reference junction and a differential amplifier. The data were processed by analog to digital conversion resulting in temperature values and temperature/time plots.

The time lapse (1 hour) between fixation of the bridge and the first recording was used for acclimatization. To avoid physiological fluctuation of the core temperature on the measurements, the tests were performed at the same time of each measurement day under normal room temperature (21-22^{o}C) conditions. Since oral breathing influences the ambient temperature of teeth, the test persons were asked to breath during the test only through their noses.

As soon as the initial ambient temperature within the oral cavity was constant for at least 5 min, the recording period of 3 min started, in which the change in temperature was registered as a result of the thermal load. The experimental data obtained were compared with the data of FEA.

RESULTS

In Figure 2 the calculated data at node S and P are shown induced by a positive thermal load having at time t = 0 a maximum temperature of 48.6^{o}C, i.e., the mean maximum recorded temperature at A and C (Fig 1). The corresponding maximum temperatures for nodes S and P (Table 1) were achieved in 0.2 sec and 25.6 sec, respectively. The minimum calculated temperatures at nodes S and P as result of the negative load (15.7^{o}C: the mean minimum temperature of A and C) are given in Table 1.

The plots with both the minimum and maximum recorded temperature values of one test person are presented in Figure 3 resulting from 1 draught of hot coffee. Notwithstanding the standardized experimental condition,

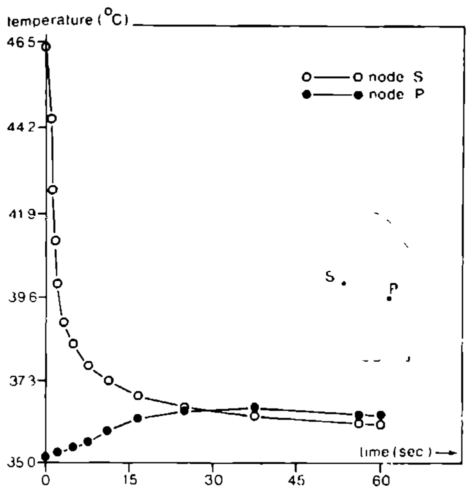


Fig 2 - Calculated temperature change at nodes P and S after initiating the heat flux.

Table 1: Theoretical data as calculated for nodes S (at surface) and P (at PDJ)

node	temperature (°C) initial	maximum	minimum
S	35.2	46.4	18.9
P	35.2	36.5	33.3

Figure 3 shows a great difference in temperature change at the measuring sites A and C, whereas the temperature change at B within the pontic appears to have less variation.

The recorded mean temperatures at A, B, and C are noted in Table 2. It is apparent that the measuring sites A and C differ in temperatures (Table 2). At A the recorded temperatures appear to be higher than at site C. The maximum and minimum temperatures were recorded in 1.3 ± 0.5 sec at A/C and 27.3 ± 4.1 sec at B after the first initial temperature change occurred.

For the positive load situation, the difference between the calculated and experimental data at the measuring site within the model was 1.0°C. The recorded initial temperatures appeared to agree with our assumption.

Apart from the sign, no significant differences were found between the temperature changes (calculated/recorded) as a result of the cold drink and those obtained for the positive thermal load.

Table 2: Mean temperatures recorded at A, B and C (see Fig 1) for two test persons (I and II)

site	test person	mean temperatures in °C (+ sd) initial n: 10	maximum 5	minimum 5	mean (I + II) initial 20	maximum 10	minimum 10
A	I	34.8(0.6)	49.8(3.1)	16.3(2.1)	35.2(0.7)	50.5(3.6)	13.4(4.0)
	II	35.7(0.5)	51.2(6.9)	10.6(3.4)			
B	I	35.4(0.3)	37.3(0.1)	33.2(0.3)	35.6(0.3)	37.5(0.2)	33.4(0.5)
	II	35.7(0.1)	37.6(0.3)	33.6(0.5)			
C	I	35.1(0.6)	46.3(2.1)	20.1(2.6)	34.9(0.6)	46.6(2.5)	18.1(3.1)
	II	34.8(0.6)	46.8(3.2)	16.0(2.3)			

DISCUSSION

The thermal diffusivity is an important parameter in transient heat transport problems (Braden 1964b). In the theoretical model, the value of thermal diffusivity used for epoxy was obtained from a laboratory study (chapter 5.2), having a standard deviation of less than 10%. In a pilot study, the thermal diffusivity was enlarged by 10%. With respect to the present calculated data for the nodes S and P, the temperatures at these nodes, given this increased thermal diffusivity, showed a deviation of 0.6% and 0.3%, respectively. In view of the recorded values at A/C and B, the value of the thermal diffusivity could vary within 10% without significantly affecting the results.

In the case of vital teeth in situ, the subgingival part of the root is insulated by the environmental tissues. For FEA in these cases, the temperature at the subgingival cervical outer surface is assumed to be prescribed (chapter 4.1). Therefore, for a pontic as tooth model, it should be taken into account that this verification model does not correspond with the biological reality.

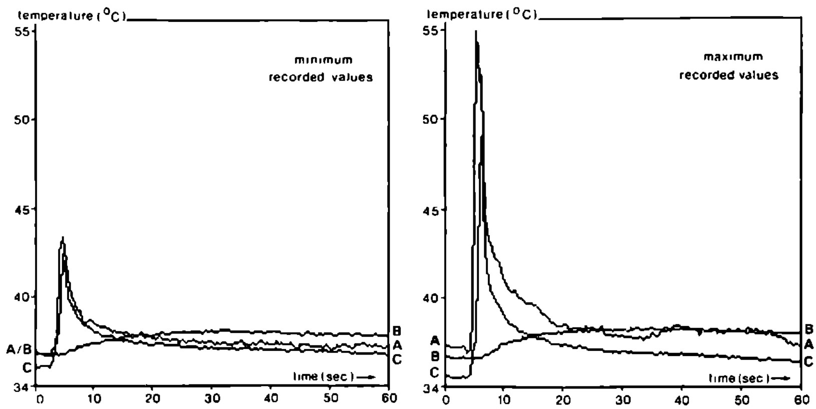


Fig 3 - Recorded temperature change at the sites A, B, and C in one person drinking a 30 ml draught of hot coffee at 60°C.

Under the given conditions, the change in ambient temperature at A and C (Fig 3) agrees with the mathematical description of the thermal load as a function of time by using an error function. The corresponding site of A/C is node S, the temperature of which has been calculated by extrapolation of the temperatures just underneath the outer surface, whereas in fact A and C do not touch the outer surface of the adjacent crowns. A direct comparison of the temperature curve at node S with those of the sites A/C is not justified without comment. In view of the small temperature differences between the sites A/C and S, it can be stated that the used value of htc causes a maximum input of heat through the surface. Since enlargement of this htc-value did not improve the reported results within the theoretical model at point P, the presented value appears to correspond with the amount of energy transported into the model to cause the given temperatures recorded at site B.

Comparing the recorded temperatures at sites A and C (Table 2), it appears that the thermal load acting on teeth in the anterior region of the oral cavity (A) is larger than in the posterior region (C). However, the difference in temperature is not significant for the recorded variation in temperatures at A and C within one and the same oral cavity (Fig 3). Moreover, this thermal behavior is not supported by data from an in vivo study using similar loading conditions (chapter 5.1). Therefore, at present no definite conclusion can be drawn with respect to the difference in temperature between anterior and posterior regions.

The results recorded at sites A/C are in agreement with the results of similar research (chapter 5.1; Plant 1974). The maximum temperature recorded at the palatal side of the upper molar tooth by Spierings (chapter 5.1) was $55.9 \pm 3.7^{o}C$ as a result of a draught of hot coffee ($65^{o}C$), whereas a cool drink ($5^{o}C$) resulted in $14.3 \pm 3.3^{o}C$. Plant (1974) recorded temperatures of $50.0-53.3^{o}C$ as a result of hot coffee ($58.0-63.5^{o}C$).

It can be concluded that, under the given conditions, the simulated model was demonstrated to be a good approximation of the physical reality. The value of htc of 5.10^{4} $J/m^{2}s^{o}C$ can be used in further studies concerning thermal analysis within restored teeth using similar loading conditions.

GENERAL DISCUSSION AND CONCLUSIONS

DISCUSSION

The influence of the size of a restoration and the properties of restorative materials on temperature distribution within a tooth has been widely recognized. To gain more insight into these phenomena, thermal analyses have been performed in a theoretical tooth model using the finite element method (chapter 4). Since this method describes the complex physical reality with a simplified mathematical model, it could be argued that in vivo experiments would provide more direct information. Thus far, however, in vivo measurements in the pulp or at the pulpo-dentinal junction (PDJ) are not possible without affecting the ambient tissues, thereby influencing the temperature to be measured. Also, the wide biological variation in geometry and material properties of teeth will seriously impair standardization of in vivo experiments. Therefore, in this thesis, a theoretical method has been used first to investigate the thermal problem in general.

Using this theoretical model, the relative importance of the various parameters involved has been studied (section 4.1). This analysis gave rise to a closer examination of the thermal load as part of the modeling. Since thermal loading conditions, involved in drinking a liquid are essential for subsequent thermal effects within teeth, these conditions were of major concern (chapter 5). Loading conditions were studied using both in vivo experiments and a combination of experimental and theoretical research. The results obtained (sections 5.1, 5.3 and 5.4) led to new input data for the theoretical model. These data indicate that the thermal load acting on teeth during the consumption of a beverage is more severe than previously assumed in our theoretical model (chapter 4).

The heat transfer coefficient (htc) determines the rate for transient temperature changes within the tooth models (section 4.1). It is therefore an important parameter in any theoretical model. The htc-value of the heat transfer coefficient of 5.10^4 J/m^2s^oC satisfies for the loading conditions under study. Increasing this value appeared to have no influence on the calculated temperatures at the top of the PDJ, whereas the temperature gradient to the surface increased (section 5.3). Under equal thermal loading, reduction of the value of this coefficient minimized the temperature changes in the entire theoretical model. Thus in this study, the value of the htc represents in fact the lower bound. Since the temperature recordings were limited to one site at the PDJ, it can be disputed whether temperatures recorded at other sites in the physical model, e.g., nearer the tooth's surface, correspond with the calculated data using this value of htc. Therefore, for other inquiries, it may be desirable to determine the extreme value of htc through experimental or theoretical research.

In future FEA studies concerning the temperature distribution within restored teeth, it is recommended that the calculations for the loading condition obtained in this thesis be carried out first and then for the extreme loading condition, i.e., the same load but with an infinitely large value of htc. From these results, the influence of the value of htc on the temperatures within the restored models might be analyzed before additional experiments are carried out.

Regarding the problems involved in in vivo experiments, the theoretical studies as described in this thesis are a first approach to a clear comprehension of temperature distribution within teeth. The thermal analyses have shown that under the given conditions, the insulating effect of cement bases to the underlying dentin is of less importance than is suggested in textbooks on operative dentistry.

In view of our results and with respect to the postoperative thermal discomfort (Piperno 1982; Miller 1984), it is still in question which criteria have to be used for a cement base to be a good insulator: (1) the temperature change at the cement base-dentin interface or at the PDJ below the cement base; (2) the temperature gradient within restored teeth; (3) the specific heat and the amount of the restorative materials used; or (4) the degree of postoperative thermal response to biological thermal loading conditions. Both the temperature change and the temperature gradient might lead to pulp damage, depending on the recuperative ability of the tissue involved (chapter 2). With respect to the temperature gradient, the degree of injury depends on the duration of the applied temperature field. Injury could also be induced by a gradual change in ambient temperature. The specific heat should be larger than $0.6\ 10^{-3}$ J/kg^{o}C (McCabe 1980). However, this cannot be used as a discriminating criterion, since in fact all cement bases meet this criterion. Moreover, the presence of a calcium hydroxide cement base underneath an amalgam restoration appeared to be insufficient to prevent postoperative discomfort (Silvestri 1977; Piperno 1982; Miller 1984). The quantity of restorative material appears to be more important than the specific heat of the material itself (chapter 4). Criterion (4), the degree of thermal response, depends on the thermal sensitivity of the nerve endings (Närhi 1985), in combination with the biological variation of the tissues, and the subjective experience of pain to thermal stimuli. All four criteria thus depend on a complex of given circumstances. Thus far, no criterion could be formulated which restorative materials should meet to protect tooth tissues against irreversible thermal injury. Therefore, extensive further research will be necessary to determine the acceptable temperature changes within teeth to prevent thermal injuries and to develop directives for dental practitioners to avoid these thermal problems.

The influence of the blood circulation in the dental pulp on temperature distribution within teeth has not been studied in this thesis but could be accounted for by using the FEA. However, a blood flow of 3.6-16.3 cm^3/s.kg of pulp tissue, as determined in a canine of a young dog (Meyer 1964; 1974), results only in a flow of $24.5\ 10^{-5}$-$5.5\ 10^{-5}$ ml/s, which might be considered as negligible compared to the quantity of surrounding tissues.

Finite element analysis also opens the opportunity to calculate the temperature distribution as well as the thermal stresses within teeth under various loading conditions. Since dental treatment, such as drilling a decayed tooth, initiates postoperative discomfort, the development of new techniques using laser beams could minimize subsequent injuries. The effect by using lasers on the temperature distribution within teeth can be calculated in a theoretical tooth model, provided the amount of energy released is known. According to Salomon (1986), no degenerative pulp tissue was manifest in teeth treated with laser beams. The highly local dissipation of energy might explain this positive result. However, it is too early to state that laser techniques for cutting cavities will not affect the tissues thermally. FEA can be helpful to answer the many open questions.

In view of the thermal and esthetic aspects of amalgam restored teeth, the use of posterior composites and glass-ionomer cements is of fast growing interest. However, the use of heat to reach optimum polymerization in light-cured composites (Braem 1986) could negatively influence the after-effects on the pulp. Therefore, it is recommended that the temperature distribution within teeth as result of such a thermal load has to be investigated first, before introducing such restorative techniques in general practice. The use of FEA can be helpful again in directing the development of these new restorative procedures.

In direct line with this thesis, other aspects may also be of interest for further research. For instance, it is still disputed whether an axisymmetric tooth model might be used for thermal analysis. Therefore, next to the verification of the theoretical model with an in vivo experiment, such as is investigated in this thesis, it is recommended that the accuracy of such a theoretical model be verified with a real three-dimensional replica of a tooth.

Another aspect concerns the heat transport at the internal interfaces, which might induce a temperature gradient between the different materials resulting in mechanical effects, such as tensil, compressive, and shear forces or even relative motion along the interface. Thus, with respect to the long-term mechanical behavior of any restoration in the oral cavity, the thermal aspects of the interfaces involved have to be taken into account as well.

CONCLUSIONS

1 The use of finite element analyses prior to experimental research of thermal problems in restored teeth, showed the need for further research into thermal loading and heat transport aspects;

2 The combination of theoretical and experimental research appeared to be effective in determining the relative importance of the various physical parameters involved in the complex effects of a thermal load acting on restored teeth;

3 The importance of the thickness of cement bases for insulating purposes, as mentioned in textbooks on operative dentistry, is over-emphasized. More important factors may be the cavity preparation procedures, the physical properties of restorative materials, and the use of cooling systems;

4 In transient heat transport problems, the heat transfer coefficient is more important than the thermal diffusivity as a rate-determining parameter for transient temperature changes;

5 The value of the heat transfer coefficient is an essential parameter within the system under study. The value of 5.10^4 J/m^2s^oC has to be taken for the given circumstances;

6 The described axisymmetric tooth model proved to be a reliable and valuable tool in fundamental studies of thermal behavior of restored teeth.

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SUMMARY

Thermal shocks in the oral cavity may affect teeth and thereby damage their structures. Temperature changes during dental treatment, and subsequent histological changes of the pulp tissue have been widely reported in the literature. However, only limited data are available concerning the influence of the presence of metallic restorative materials and the presence of cement bases on the temperature distribution within teeth. The aim of the present investigation is to analyze the heat transmission and temperature distribution within various restored teeth under certain transient thermal loading conditions.

In chapter 2 a literature review is given concerning the thermal aspects in the oral cavity. Based on data from the literature, the surface temperatures for soft tissues and teeth are discussed and summarized in section 2.1. Section 2.2 deals with various thermal loading conditions of which the after-effects on tooth structures are reviewed. It is concluded that hardly any information is available concerning the thermal distribution and the thermal processes in human teeth. One of the described after-effects subsequent to restorative procedures is postoperative discomfort like thermal hypersensitivity which occurs in 38 to 50% of the cases. Some recommendations are given to minimize postoperative discomfort.

To gain insight into the thermal behavior of teeth, a mathematical model of a molar tooth has been developed simulating different thermal processes by simple parameter variation. This theoretical simulation model is described in chapter 3. Thermal analyses have been carried out using Finite Element Analysis (FEA). The calculated temperatures appeared to be in good agreement with clinical experimental research reported in the literature. It is concluded that the model is a valid tool for further research with respect to the influence of restorative materials and cavity geometry on thermal behavior of restored teeth.

In chapter 4, the influence of geometry and restorative materials on the temperature field was analyzed in an axisymmetric tooth model using FEA. The models of an unrestored tooth and different restored teeth were evaluated comparatively. The insulating ability of a calcium hydroxide cement base is low under the given conditions, due to its insufficient thermal properties. The use of a double base, i.e., 0.5 mm calcium hydroxide plus 1.5 mm polymer-modified zinc oxide-eugenol, could be recommended to avoid thermal discomfort in the postoperative phase. The influence of cavity geometry on the direction of the heat flow is negligible in the vicinity of the pulp and restoration. However, with respect to a standard occlusal restoration, an amalgam build-up showed an increase in the maximum temperature as well as in the rate of the temperature rise at the pulpo-dentinal junction. The temperature field within the restored tooth depends not only on the size of the restoration but also on the properties of the restorative materials used.

Parametric studies related to the thermal conductivity of amalgam has indicated that an increased value of the thermal conductivity of amalgam does not influence the calculations on temperature distribution within the model. The convective heat transfer coefficient (htc), however, turned out to be an essential parameter. This coefficient represents the quantity of thermal energy transferred in a unit time at a fluid-solid interface of a unit area which has a unit temperature difference. Lack of sufficient information about the real value of the htc, and the demonstrated importance of this parameter when analyzing thermal problems gave rise to additional experimental research.

In the chosen model some assumptions have been made concerning the

thermal load in the oral cavity during drinking hot/cool liquids. Therefore, in section 5.1 attention is firstly focused on an in vivo experiment. Based on the recorded ambient temperatures of teeth, an error function is found to comply with all recorded temperatures.

In section 5.2, a polyester and an epoxy resin were examined on their value of thermal diffusivity and on their suitability for pouring replicas. The polyester resin proved to be unsuitable for this purpose, whereas the epoxy resin can be used as replica resin. A mixture of this resin was molded into cubic specimens and into the replica's used in the following experiments. The cubic specimens were used for direct analysis of the thermal diffusivity. The mean value of the thermal diffusivity of the mixture was $1.40 \pm 0.05\ 10^{-7}\ m^2/s$.

To determine the value of the htc by means of only laboratory experiments is rather difficult. Therefore, in section 5.3, a computer aided method is developed. The different loading conditions were exerted in theoretical simulation models as well as in laboratory experiments. Comparison of the theoretical data with the experimental results led to the conclusion that the htc-value of $5.10^4\ J/m^2s^oC$ satisfies for the experimental conditions used.

Section 5.4 deals with the findings of additional in vivo experiments concerning the value of the htc during and after the consumption of liquids. The recorded data obtained were compared with the data of FEA as result of the mathematical tooth model. The htc-value found in section 5.3 can be used in further studies concerning thermal analysis with restored teeth using similar loading conditions. Finally it can be concluded that under the given conditions the simulated tooth model appeared to be a good approximation of the physical reality.

SAMENVATTING

Grote temperatuurveranderingen kunnen in de mondholte aanleiding geven tot beschadiging van de dentitie. In de literatuur is de aandacht vooral gericht op de temperatuurveranderingen die optreden in de dentitie tijdens een tandheelkundige behandeling en op de hieruit voortvloeiende histologische veranderingen van het pulpa-weefsel. Over het thermisch gedrag van een gebitselement en de invloed van metalen restauratiematerialen of onderlagen op het temperatuurbeeld ten gevolge van een thermische belasting is echter weinig bekend. De doelstelling van het onderhavige onderzoek was om in - op verschillende wijze - gerestaureerde gebitselementen het warmte-transport en de temperatuurverdeling te analyseren als gevolg van een tijdsafhankelijke thermische belasting.

Hoofdstuk 2 geeft een literatuuroverzicht met betrekking tot de thermische aspecten in de mondholte in het algemeen en tot die van het gebitselement in het bijzonder. Op grond van dit literatuuronderzoek worden in paragraaf 2.1 de oppervlakte temperaturen van de zachte weefsels en die van de dentitie besproken en samengevat.

In paragraaf 2.2 wordt verslag gedaan van verschillende thermische belastingscondities op tandstructuren en de mogelijke gevolgen daarvan. Uit dit overzicht blijkt ondermeer dat er bijna geen informatie beschikbaar is over de thermische processen en de temperatuurverdeling in menselijke gebitselementen. Wel is bekend dat het post-operatieve ongemak na een restauratieve ingreep, zoals verhoogde thermische gevoeligheid, in 38 tot 50% van de gevallen voorkomt. Ter afsluiting van dit hoofdstuk worden nog enkele aanbevelingen gegeven om post-operatieve klachten tot een minimum te beperken.

Teneinde inzicht te krijgen in het thermisch gedrag van gebitselementen is gebruik gemaakt van een mathematische methode. Hiertoe is een theoretisch model van een molaar ontwikkeld, dat door middel van eenvoudige parametervariaties verschillende thermische processen kan simuleren. Dit theoretische simulatiemodel is beschreven in hoofdstuk 3. Voor het uitvoeren van de thermische analyses werd gebruik gemaakt van de Eindige Elementen Methode (EEM). De hiermee berekende temperaturen bleken overeen te komen met resultaten van klinisch experimenteel onderzoek uit de literatuur. Het model blijkt een valide hulpmiddel voor verder onderzoek naar de invloed van restauratiematerialen en caviteitsgeometrie op het temperatuurbeeld van gerestaureerde gebitselementen.

Hoofdstuk 4 gaat nader in op de hierboven genoemde invloeden op het temperatuurbeeld in een axisymmetrisch tandmodel als gevolg van een thermische belasting en met gebruikmaking van de EEM. De geometrisch identieke modellen, die geen dan wel verschillende restauraties bevatten, werden thermisch belast. De verkregen temperatuurbeelden in de modellen werden met elkaar vergeleken. Onder de gegeven condities blijkt de isolerende kwaliteit van een calciumhydroxide onderlaag gering te zijn als gevolg van de onvoldoende thermische eigenschappen. Overwogen kan worden om een dubbele onderlaag aan te brengen (bv. 0.5 mm calciumhydroxide plus 1.5 mm gemodificeerde zinkoxide-eugenol) teneinde in de post-operatieve fase het thermische ongemak te beperken.

Uit dit onderzoek is gebleken dat de richting van de warmtestroom in de omgeving van de pulpa niet beinvloed wordt door de caviteitsgeometrie. De snelheid waarmee de temperatuur toeneemt alsmede de berekende maximum temperatuur op de overgang dentine-pulpa zijn echter aanzienlijk hoger in een amalgaam kroon dan in een standaard occlusale amalgaamrestauratie. Het temperatuurbeeld in een gerestaureerd gebitselement is zowel afhankelijk van de thermische eigenschappen van het toegepaste restauratiemateriaal

als van de omvang van de restauratie.

Naast bovengenoemde thermische analyses is een parameterstudie uitgevoerd. Hieruit volgde dat met name de warmteoverdrachtscoefficient (woc) het temperatuurbeeld in de modellen sterk beinvloedde. Onvoldoende informatie omtrent de werkelijke waarde van deze parameter leidde derhalve tot aanvullend experimenteel onderzoek.

De thermische belasting gedurende het drinken van hete/gekoelde vloeistoffen is in het onderhavige model in eerste instantie gebaseerd op een schatting en wordt in paragraaf 5.1 in een in vivo experiment nader bepaald. Op grond van de verkregen resultaten kan de temperatuur-tijd curve van de omgevingstemperatuur van de gebitselementen ten gevolge van het drinken van een slok warme/koude vloeistof mathematisch beschreven worden als een error-functie.

In paragraaf 5.2 worden van een polyesterhars en een epoxyhars de thermische diffusie en de verwerkingsprocedure onderzocht op hun geschiktheid om te dienen als kunststoffen replicas van gebitselementen. Het polyesterkunsthars bleek voor dit doel ongeschikt. Van het epoxyhars konden echter wel op eenvoudige wijze replicas worden vervaardigd. Uit een mengsel van dit epoxyhars werden zowel de replicas voor het vervolg onderzoek gegoten alsmede de kubusvormige testmonsters. De testmonsters werden gebruikt voor rechtstreekse analyse van de thermische diffusie van het epoxyhars. Een gemiddelde waarde van de thermisch diffusie van $1.40 \pm 0.05\ 10^{-7}\ m^2/s$ werd gevonden.

Het bepalen van de waarde van de woc in een laboratoriumopstelling is moeilijk. Derhalve is in paragraaf 5.3 gebruik gemaakt van een methode die door een computerprogramma wordt ondersteund. De verschillende belastingcondities werden hiertoe gesimuleerd in een theoretisch en in een experimenteel model. Vergelijking van de theoretische data met de experimentele gegevens resulteerde in een woc-waarde van $5.10^4\ J/m^2s^oC$.

Op grond van de bevindingen in paragraaf 5.3 werd in paragraaf 5.4 een aanvullend in vivo onderzoek verricht ter nadere bepaling van de woc-waarde tijdens het consumeren van warme/koude dranken. De verkregen experimentele gegevens vertoonden goede overeenkomsten met die van het theoretisch simulatiemodel. De in paragraaf 5.3 gevonden woc-waarde voldoet en kan worden toegepast in toekomstige studies die gericht zijn op thermische analyse van gerestaureerde gebitselementen. Hierbij dient wel voldaan te zijn aan de voorwaarde dat soortgelijke thermische condities zich voordoen aan het oppervlak van het gebitselement. Tenslotte kan worden vastgesteld dat, rekening houdend met de gegeven condities, het simulatiemodel een goede benadering is van de fysische realiteit.

CURRICULUM VITAE

De schrijfster van dit proefschrift werd 19 mei 1952 geboren te 's-Hertogenbosch. Na het behalen van de diploma's MULO-B (mei 1968) en HBS-B (mei 1972) begon zij in 1972 met de studie tandheelkunde te Nijmegen. In 1978 behaalde zij het doctoraalexamen en in mei van hetzelfde jaar het tandartsexamen.

Van 1978 tot 1980 was zij als hoofdinstructeur half-time verbonden aan de afdeling Conserverende Tandheelkunde voor Volwassenen te Nijmegen (hoofd: Prof. Dr. A.J.M. Plasschaert) en part-time werkzaam in een algemene tandartsenpraktijk te Millingen a/d Rijn. In 1980 werd zij wetenschappelijk medewerker (8/10 weektaak) binnen de hierboven genoemde afdeling, waar het hier beschreven onderzoek werd verricht.

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STELLINGEN

1. Ten behoeve van het optimaal comfort voor bewoners in goed geïsoleerde huizen verdient het aanbeveling de huidige verwarmingsinstallatieberekeningen te corrigeren op de thermisch dynamische gebouweigenschappen. (DIN 4701; ISSO-4)

2. De veronderstelling dat de oorzakelijke faktor van mondkanker ondermeer gelegen zou zijn in een thermische belasting van de mondslijmvliezen tijdens het roken is onjuist. (Fritz: Quintessenz, 1984)

3. Het belang van een cementonderlaag als thermische isolator bij het restaureren van gebitselementen wordt overschat. (dit proefschrift)

4. Bij het introduceren van het begrip "Specific Grinding Energy" (Westland: Ned Tijdsch Tandh, 1978) om het rendement van preparatietechnieken te kunnen kwantificeren is geen rekening gehouden met de thermische gevoeligheid van gebitselementen.

5. Om esthetisch redenen is in frontelementen Durelon^{R} als onderlaag te prefereren boven Dycal^{R}. (vd Burgt: Oral Surg 1985)

6. Bij het vroegtijdig onderkennen van het Pierre Robin syndroom mag men de ogen niet sluiten voor het tot blindheid leidende Stickler syndroom. (Saksena: J Craniofac Genet Dev Biol, 1983)

7. Vanuit sociaal oogpunt is het aan te bevelen bij kinderen met een afwijkend habitueel mondgedrag, waarbij logopedische therapie is geïndiceerd, deze therapie uit te voeren voordat een eventuele orthodontische behandeling wordt begonnen.

8. Om de hoeveelheid vetopslag bij een vrouw gedurende de zwangerschap niet te overschatten, dient correctie plaats te vinden voor het soortelijk gewicht van de aanwezige vetvrije massa. (Schonk: Annual Report Nestlé Foundation, 1985)

9. Het beschermen van bedreigde dier- en plantensoorten heeft alleen zin, wanneer tegelijkertijd hun natuurlijk milieu wordt beschermd.

10. Omdat atoomsplitsing alles heeft veranderd, behalve onze manier van denken, is het niet denkbeeldig dat we op een onvoorstelbare ramp aansturen. (Albert Einstein)

11. Elk weer waar je met je schip in vaart nadat je aan lager wal bent ontsnapt, is heel plezierig, hoewel sommig weer plezieriger is dan ander. (Joshua Slogan)

12. Het is billijk de universitaire bijdrage voor promotiekosten te relateren aan het inkomensnivo van de promovendus. Hierbij kan de grootte van de bijdrage gekoppeld worden aan de omvang van de sociale verplichtingen én aan het toekomstperspectief van de jonge doctor.

Stellingen behorend bij het proefschrift: Modeling and analysis of thermal loading and heat transport in restored teeth.
Doret Spierings, 30 mei 1986.